A Comparison of Biomechanical Variables, Neuromuscular Control and Strength during Controlled and Unexpected Falls on the Outstretched Hands in Young and Older Women

A Thesis Submitted to the College of
Graduate Studies and Research
In Partial Fulfillment of the Requirements
For the Degree of Doctor of Philosophy
In the College of Kinesiology
University of Saskatchewan
Saskatoon

By

LAUREN JANE LATTIMER

PERMISSION TO USE

In presenting this thesis/dissertation in partial fulfillment of the requirements for a Postgraduate degree from the University of Saskatchewan, I agree that the Libraries of this University may make it freely available for inspection. I further agree that permission for copying of this thesis/dissertation in any manner, in whole or in part, for scholarly purposes may be granted by the professor or professors who supervised my thesis/dissertation work or, in their absence, by the Head of the Department or the Dean of the College in which my thesis work was done. It is understood that any copying or publication or use of this thesis/dissertation or parts thereof for financial gain shall not be allowed without my written permission. It is also understood that due recognition shall be given to me and to the University of Saskatchewan in any scholarly use which may be made of any material in my thesis/dissertation. Requests for permission to copy or to make other uses of materials in this thesis/dissertation in whole or part should be addressed to:

Dr. Carol Rodgers
College of Kinesiology
University of Saskatchewan
87 Campus Drive
Saskatoon, Saskatchewan, Canada S7N 5B2

OR

Dr. Adam Baxter-Jones
College of Graduate Studies and Research
University of Saskatchewan
107 Administration Place
Saskatoon, Saskatchewan, Canada S7N 5A2

ABSTRACT

Purpose: This thesis evaluated the age differences in biomechanics and muscle activity during controlled and unexpected descents simulating a fall on the outstretched hands (FOOSH) in women. Laboratory simulation using two different protocols investigated this common mechanism of injury in older and younger women. The primary purpose of the controlled descent (FOOSH 1) was to examine the differences between young and older women to control the post-impact phase of a forward fall descent at three body angles. The primary purpose of the unexpected descent (FOOSH 2) was to examine biomechanical and muscle activity age differences in pre-impact, impact and post-impact phases of a simulated FOOSH.

Methods: FOOSH 1 was a cross sectional study comparing twenty healthy young (mean 24.8±3.4 yrs.) and 18 healthy older (68.4±5.7 yrs.) women performing controlled descents on outstretched arms at three body lean angles (60, 45, and 30° from horizontal) and a muscle strength test of the non-dominant UE [isometric (ISO) concentric (CON) and eccentric (ECC)] using an isokinetic dynamometer. FOOSH 2, also a cross sectional design, evaluated twenty young (mean age 22.9 yrs., SD±3.7) and 16 older (mean age 68.1yrs., SD ±5.0) women performing five trials of unexpected FOOSHs at a body lean angle of 60° from horizontal with the same muscle strength testing protocol. A three-dimensional motion capture system (VICON Nexus, VICON, Centennial, CO) and force plate apparatus (OR6-7, AMTI, Watertown, MA) was used to determine the biomechanical measures of peak energy absorption, maximum vertical force, maximum elbow angle and maximum elbow joint extensor moment. Additional biomechanical measures of FOOSH 2 included: elbow angle and elbow angular velocity at impact, elbow joint stiffness, end elbow angle, and impulse. Surface EMG detected muscle activity of six muscle sites: anterior deltoid (AntDEL), pectoralis major (PM), triceps brachii (long head) (TRI), biceps brachii (BB), external oblique (EO) and internal oblique/transversus abdominus (IO/TrA).

Results: In FOOSH 1 and FOOSH 2, older women demonstrated decreased CON elbow extensor strength compared with younger women. During FOOSH 1, at all angles, the older women had increased BB activity and decreased EO activity. In FOOSH 2 older women had significantly less IO/TrA activity prior to impact than younger women. The women differed in landing strategy in that younger women had significantly greater elbow joint angle and velocity at impact. Older women demonstrated diminished capacity to absorb energy in both the controlled (30°) and unexpected descent. **Significance of findings:** This is the first study to investigate biomechanical and muscle activation age differences for a simulated controlled and unexpected forward descent in women. Older women demonstrate differences that could potentially increase their risk of injury during a forward fall. The results of these studies could help clinicians develop fall injury prevention protocols by considering the neuromuscular and biomechanical factors that are important to control a forward descent. The findings suggest that UE and trunk muscle strengthening may be important components to include in a fall injury prevention training program. The modulation of energy absorption capabilities by altering elbow velocity and increasing elbow flexion angles at impact may also be an injury prevention tactic to be adopted.

PREFACE AND AUTHOR CONTRIBUTIONS

I, Lauren Lattimer, was the primary author of all chapters within this thesis. Chapters 3 to 5 represent manuscripts that have either been submitted to or have been published in peer-reviewed journals. Author contributions have been discussed and approved by the student advisory committee and were included as part of the student-supervisor agreement. Chapter 3 represents a manuscript that has been prepared and submitted to Journal of Aging and Physical Activity and was co-authored by Dr. Joel Lanovaz, Dr. Jonathan Farthing, Dr. Stephanie Madill, Dr. Soo Kim, Dr. Stephen Robinovitch, and Dr. Catherine Arnold. Dr. Catherine Arnold and Dr. Joel Lanovaz developed the initial laboratory testing protocol and I further refined and pilot tested the protocol with the assistance of a group of Masters of Physical Therapy students. Under the guidance of Dr. Jonathan Farthing, I developed and pilot tested the strength testing protocol. I conducted the data collection, data processing, data analysis, interpreted results and prepared the manuscript under supervision of my supervisor, Dr. Arnold, and the authors mentioned previously. A group of Physical Therapy students; Erin Gibb, Anastasia Slobodzian, Matthew Ankerrman, Keenan Oberg, Leah Sauchyn aided in protocol development, and initial data collection and presented a separate poster presentation and assignment that had no overlap with my project. Chapter 4 represents a manuscript that has been accepted for publication in the Journal of Electromyography and Kinesiology. Chapter 4 was co-authored by Dr. Joel Lanovaz, Dr. Jonathan Farthing, Dr. Stephanie Madill, Dr. Soo Kim, and Dr. Catherine Arnold. I contributed to the laboratory design and conducted the data collection, data processing, and interpretation. The manuscript was prepared under the guidance of the above authors. Chapter 5 represents a manuscript that will be submitted to Clinical Biomechanics. Chapter 5 utilized the same subjects as Chapter 4, but focused on different research questions and variables measured. The manuscript was prepared under the guidance of my supervisor, Dr. Arnold, and Dr. Joel

Lanovaz, Dr. Jonathan Farthing, Dr. Stephanie Madill, Dr. Soo Kim, and Dr. Stephen Robinovitch. Dr. Lanovaz provided help with the preparation of the data including MATLAB computer scripts, and provided laboratory support throughout the data collection process for all three manuscripts.

ACKNOWLEDGMENTS

I would like to acknowledge my supervisor Dr. Cathy Arnold in her integral role in completion of this thesis and my personal development as a researcher. She remained incredibly supportive and provided excellent guidance in all aspects of the process towards completion. Secondly, I would like to acknowledge my advisory committee, whose vast knowledge and expertise was invaluable. I recognize the time and investment put forth by my supervisor and committee and could not be more appreciative. Collaboration with this group of academics has been the most enriching part of my graduate degree. The supportive mentorship has been incredibly impactful and will continue to be throughout my career.

This work was supported by The Schulman Foundation, College of Medicine, University of Saskatchewan and the Physiotherapy Foundation of Canada. These funding sources had no involvement in the study design, collection, analysis or interpretation of the data.

Acknowledgements

I would also like to acknowledge the following graduate and undergraduate students who contributed to aspects of design and data collection: Tanner Treen, Janelle Unger, Joel Krentz, Isabel Hedayet, Braden Ripley, Kyle Zurevinsky. Erin Gibb, Anastasia Slobodzian, Matthew Ankerrman, Keenan Oberg and Leah Sauchyn.

LIST OF TABLES

LIST OF FIGURES

CONCEPTUAL DEFINITIONS

Fall: an event which results in a person coming to rest inadvertently on the ground or floor or other lower level [1]

Isotonic: A constant force or resistance [2]

Muscle action: The development of muscle tension [3]

Eccentric contraction: An isotonic muscle action in which the muscle lengthens as it contracts [3]

Concentric contraction: An isotonic muscle action in which the muscle shortens as it contracts [3]

Isometric contraction: A muscle action in which the muscle does not change length [2]

Isokinetic contraction: The application of force at a constant speed against an equal force [2]

Isokinetic dynamometry: Measurement of muscular force with a constant velocity [2]

Electromyography: electrical recording of muscle activity [4]

Muscle force: the force exerted by structural and active elements of a muscle [2]

Joint moment: the resultant muscle force exerted by a group of muscles about a joint [3]

Range of motion: the maximum angular displacement about a joint [3]

Agonist: A muscles whose activation produces the acceleration required for a movement [2]

Antagonist: A muscles whose activation produces an acceleration in the direction opposite to that required for a movement [2]

LIST OF COMMON ABBREVIATIONS USED

FOOSH	Fall on outstretched hand
EMG	Electromyography
WHQ	Waterloo handedness questionnaire
MVC	Maximum voluntary isometric contraction
AntDEL	Anterior deltoid muscle
PEC	Pectoralis major muscle
BB	Biceps brachii muscle
TRI	Triceps brachii muscle (long head)
EO	External oblique muscle
IO/TrA	Internal oblique muscle/ transversus abdominus muscle
ECC	Eccentric muscle contraction
CON	Concentric muscle contraction
ISO	Isometric muscle contraction
ENRG	Energy absorption
ImV	Impact angular velocity
ImA	Impact elbow angle
ES	Elbow joint stiffness
EnA	End elbow angle
IMP	Impulse
BL	Muscle activity 500 ms prior to cable release
PRE	Muscle activity during time between cable release and impact
POST	Muscle activity during the 200ms after impact detected using force plates
Co-acPRE	The quotient of muscle activity of the BB/TRI 100ms prior to impact
Co-acPOST	The quotient of muscle activity of the BB/TRI 200ms post impact

LIST OF APPENDICES

TABLE OF CONTENTS

CHAPTER 1

INTRODUCTION

The cost of injury in Canada has increased by 35% since 2004, with predictions of increasing annual costs reaching $75 billion by 2035, an increase of 180% [5]. Falls were the leading cause of overall injury costs in Canada in 2010, accounting for $8.7 billion or 32% of total costs [5]. Fall-related injury is the leading cause of death following emergency department visits in adults over the age of 65 years [6]. An American longitudinal study found that more than one third of older adults who visited the emergency department after a fall had revisited the emergency department or died within one year [6]. In Canada, falls account for 85% of all injury related hospitalizations among older adults and the average length of stay is 21 days which is three times more than the average hospital stays among all ages. Fall- related injuries are a current and emergent major concern for older adults, health care policy makers, and the general population in Canada. Modifiable risk factors linking to the mechanism of fall- related injuries must be investigated to guide health care professionals in designing injury prevention programs.

The injuries sustained from falling can be devastating to older adults, their families, and the health care system. In Canada, there was a 65% increase in deaths due to falls from 2003 to 2008 [7]. One third of seniors in Canada who are admitted to a hospital after a fall are later discharged to long term care [8]. Falls are the cause of 95% of hip fractures and 20% of older adults who sustain a hip fracture die within one year post-fracture [7]. Head impact is common in forward falls; traumatic brain injury accounts for 32% of hospital admissions and more than 50% of the deaths from falls in older adults [9,10]. Adults who have sustained a fall over the age of 65 years account for more than 60% of the hospital admissions for traumatic brain injuries. Osteoporotic (fragility) fractures are defined as fractures sustained as the result of minimal trauma most often occurring from a fall from standing height or less [11]. Osteoporotic fractures are deemed a worldwide epidemic and are forecasted to increase with the predicted increase in population of those over 65 years [12]. A recent report indicated that over 50% of injuries due to falls occur in the upper body [7]. Fractures of the upper extremity, particularly the wrist, are common osteoporotic fractures of older adults and are more common in women [11]. Distal forearm (wrist) fractures in women typically are the result of a fall onto an outstretched hand [13].

Gender differences in fall prevalence and related injury have been observed in a large sample Canadian study where both were significantly higher in women than men [14]. Another study found fall related injury rates for all body parts are 40-60% higher in women than men of the same age [15]. The same study found that hospitalization rates were 81% higher in women compared to men [15]. This evidence suggests that women fall more often and sustain more severe injuries when they fall. Some of the potential risk factor differences to explain this disparity could be bone and muscle strength differences and the greater impact of multiple medical conditions in women than men [14].

Higher physical activity levels are protective in decreasing fall risk for both women and men [14]. Decreases in physical activity is an indicator of fall risk [16]and could further increase the risk of fall-related injury [17]. Additionally, physical activity levels can rapidly decline after a fall resulting in deterioration of functional status, social involvement and quality of life. Studies have indicated that physical activity is decreased by 15-42% in older adults after a fall [18]. This is an unfortunate outcome, as seniors who are physically active are less likely to sustain future falls [16].

The risk of sustaining a fall-related injury is complex and may be explained by several factors including individual risk factors (strength, balance, reaction time) [19]; but also other fall characteristics including the activity during the fall, fall direction, fall descent and factors occurring at fall impact such as the amount of energy absorption [20]. Aside from bone strength itself, the fall mechanism, impact forces and energy absorption of the soft tissues are critical determinants of osteoporotic fracture risk [21]. During a forward fall the potential for injury can be predicted mathematically by the resultant peak forces and moments. Counteracting the forces and moments might be modulated by factors such as neuromuscular capacity [21-23]. Evidence of ineffective use of the upper extremity (UE) has been observed to have an inverse relationship between triceps strength and hip fractures in older adults [21,23].

Skeletal muscle mass decreases by 30-50% between the ages of 40 and 80 years contributing to age-related muscle weakness [24]. This sarcopenia is attributed to many factors including decreased physical activity and alterations in hormonal status such as menopause [25]. Post-menopausal women tend to experience greater strength declines, decreased functional capacity, impairments in muscle repair and increased rates of sarcopenia with age than men [26,27]. In adults over 50 years of age muscle strength decreases at a rate of one and a half to three percent per year [28]. The loss of strength with increasing age, particularly in women, could contribute to decreases in neuromuscular function and performance during a FOOSH. For example, older women have been found to have significantly reduced ability to absorb the energy of a forward fall on the outstretched arms, simulated by an eccentric lowering task, compared to younger women [29]. Neuromuscular capacity could be an important determinant in improving energy absorption and thus preventing a serious injury during a fall, but no studies have specifically measured this potential relationship in older women.

The previous literature has focused on factors impacting fall risk and the prevention of falls. There has been little focus on understanding the biomechanical and neuromuscular demands on safely arresting a fall when the fall is inevitable [14,30-39]. Biomechanical studies of fall arrest have typically focused on a young male cohort while women over age 65 years remain the more vulnerable population to sustain fall related injury [22,40-42]. Given the devastating effects of falls on the health and well-being of older women, the strain and demands on the healthcare system, and the predicted increase in the population of adults over the age of 65 years, the need for evidence based fall injury prevention strategies becomes obvious. Examining the mechanisms of injury during a fall is a preliminary step towards guiding injury prevention protocols. There is a lack of research examining the age-related differences in neuromuscular and biomechanical performance during simulated forward falls.

This thesis is comprised of three studies, which will be discussed separately in Chapters 3 to 5 addressing important objectives to determine the neuromuscular and biomechanical age related differences that could impact injury risk in a FOOSH for older women.

Study 1 Title: Differences between young and older women in energy absorption and muscle activity during descents on the outstretched arms (FOOSH 1)

The purpose of this project was to investigate the factors associated with successful descent during a controlled FOOSH at three different body angles in younger and older women. The specific objectives were to:

1. Evaluate the differences in upper extremity (UE) concentric, isometric and eccentric strength between older and younger women;
2. Compare differences in UE and trunk muscle activity during a controlled descent on outstretched hands at different body lean angles between young and older women, and

3. Compare maximal elbow angle, elbow joint moment, contact force, and energy absorption in the non-dominant arm during a controlled descent on the outstretched arms at different body lean angles.

Hypotheses: The older women would have decreased concentric, isometric, and eccentric strength, lower trunk muscle activation, and decreased maximal elbow angle, energy absorption, elbow joint moment and increased force under the hand compared to the younger women.

Study 2 Title: Upper limb and trunk muscle activation during an unexpected descent on the outstretched hands in older and younger women (FOOSH 2)

The purpose of this study was to investigate upper extremity and trunk muscle activity during three phases: baseline (BL), pre-impact (PRE) and post-impact (POST), of an unexpected FOOSH in healthy young and older women.

Hypotheses: There will be differences in muscle activation between the young and older women. It was hypothesized that young women would have greater muscle activation amplitudes in the upper limbs and trunk prior to impact compared with the older women.

Study 3 Title: Comparison of muscle strength and biomechanics during and unexpected release on outstretched arms between older and younger women (FOOSH 2)

The purpose of this project was to investigate the factors associated with succesful landing and descent during an unexpected release and simulated FOOSH in younger and older women. Specifically the objectives were to:

1. Compare concentric, isometric and eccentric strength between the groups;
2. Compare elbow joint stiffness, angle, velocity, energy absorption and muscle co-contraction between age groups during a unexpected descent on outstretched hands, and

3. Evaluate the relationship of muscle strength with biomechanical outcomes related to injury risk

Hypotheses: Older women would have decreased concentric, isometric and eccentric strength in their upper extremities compared to younger women. Older women would have increased co-contraction during the fall arrest, land with greater impact forces, and would have increased elbow stiffness and maximum elbow flexion during an unexpected release on outstretched hands compared to younger women. Eccentric strength would positively correlate with energy absorption in both age groups.

CHAPTER 2

REVIEW OF THE LITERATURE

2.1 Forward Falls on the Outstretched Hands

2.1.1 Incidence and cost

A fall is defined as an event in which an individual unintentionally comes to rest on the ground or at a lower level [43]. Falls are the leading cause of unintentional injury across all age groups in Canada and accounted for $6.7 billion or 42% of direct costs of injury in 2010 [5,7]. Every year one third of adults over the age of 65 will sustain a fall. Falls have considerable negative physical, psychological and emotional impact on older adults and are a costly burden on the health care system. In Canada, 85% of all injury related hospitalizations are the results of falls and the average cost per fall requiring hospitalization is $29,373.00 [44]. In Canada, 67% of adults who sustain a fall visit the emergency room [7]. Fall related hospitalizations among seniors in residential care increased by 19% between 2006 and 2010 [7]. Older adults hospitalized because of a fall spend approximately three weeks in the hospital, which is three times longer than the average hospital stay in Canada [7]. Age related declines in bone density, muscle mass, and neuromuscular reflexes could amplify the severity of fall-related injury and the occurrence of falls. Most injuries to older adults result from falls, specifically 87% of fractures are the result of falls [45]. Seventy-nine percent of distal forearm fractures, in women over age 50 years, are the result of falls. After the age of 50, 4 in 10 women can expect to have a hip, vertebral or forearm fracture in their remaining lifetime and falls are the leading cause [46]. Traumatic brain injury accounts for 32% of hospital admissions and more than 50% of fall related deaths in older adults [9,10].

2.1.2 Incidence and injury risk of forward falls

Every year one third of adults over the age of 65 years will fall, 60% of which are forward falls [47]. Falling on outstretched hands (FOOSH), a protective mechanism to arrest the body and avoid injury to the head, trunk or hip, has been observed in 74% of falls videotaped in long term care [48]. More than 97 % of fractures to the upper arm are the result of a fall [12]. Forward falls and backwards are the most common mechanism of distal radius fractures in women [21]. In Canada more than 95% of hip fractures are caused by falls and 25% of adults who experience a hip fracture die within a year [7]. A recent study evaluating video data of actual falls in long term care found that the risk of hip impact in forward falls was similar to that in sideways falls [49]. Hand impact is associated with a reduced risk of hip fractures during falls and an ineffective upper extremity protective response has been positively related to hip fractures and head impact incidence in older adults [21]. An increased probability of head impact was associated with forward falls verses backward falls in a video surveillance study of older adult falls captured in the long term care setting [48]. Traumatic brain injury accounts for 32% of hospital admissions and more than 50% of fall related deaths in older adults [9,10]. Falls in young adults rarely results in head impact likely because of a combination of factors, including effective protective responses such as use of upper limbs to stop the fall, and trunk flexion and rotation during the decent [50,51].

2.1.3 Biomechanics of forward falls

The ability to successfully control a forward fall and prevent subsequent head or torso impact depends on the ability to move the hands into a protective position in order to absorb the maximum amount of energy through upper body musculature via elbow and shoulder flexion [23]. During a forward fall, the peak forces and moments, along with the counteracting resistance produced by the soft tissues, mathematically predict the potential for injury. The capability of upper extremity (UE) biomechanics to arrest falls has been studied by quantifying the impact forces of falls from minimal heights (2-5 cm) [40,52]. These impact forces are characterized by high frequency initial peak force, followed by a lower frequency peak [52]. The initial peak force that occurs shortly after impact governs fracture risk [52]. Impact forces of 1.1-4.1 kN is ample to produce a wrist fracture [53]. Falls from standing height greater than 0.6 meters can far exceed this threshold, surpassing the average fracture force of the elderly distal radius [42,52].

2.1.3.1 Phases of a forward fall arrest

Forward fall arrest has been described as having three phases: pre impact, impact and post impact descent [20] (Figure 2.1). These phases of a FOOSH are important to compartmentalize in order to highlight possible age related changes in fall arrest strategy. The pre impact phase is defined as the time period from the instant of loss of balance to hand impact on a surface [20,50]. The pre impact phase has been proposed as the phase when the neuromuscular system has the best advantage to position the body in a manner to reduce the risk of fracture on impact [20]. The pre impact phase involves rapid UE movement to prepare for impact. This movement both places the hands into position to receive the impact and prepares the UE musculature to absorb the impact forces and control the post-impact descent [54]. During falls from standing height, the time between loss of balance and hand contact averages 680± 116ms [50].

The short duration of the impact phase, defined as the time of actual impact of the distal extremity, combined with the forces on the impact limb peaking within milliseconds after impact, would suggest that neuromuscular reflexes are not fast enough to substantially modulate the reaction forces during the impact phase [20]. The impact force has been observed to peak 10-20 ms [52,55] after impact.

The post impact descent phase begins with the impact and ends with the termination of the body's forward momentum. Post impact muscle activity contributes to the deceleration and stabilization of body posture [54]. Modifications in muscle activity after hand impact do not have the potential to reduce the peak impact force on the distal forearm [40] but may play a role in diminishing the risk of impact to the head. Stretch reflexes are also likely to contribute to the muscle activity during the post impact phase [54]. Stiffness and dampening properties of the UE contribute to the energy absorbing capacity and injury probability during the post impact phase [20]. Specifically, the stiffness of the elbow joint is defined as the relationship between the deformation of a body and a given force and is calculated as the slope of the line through the moment/ angle curve and will be discussed further in 2.3.3.1.

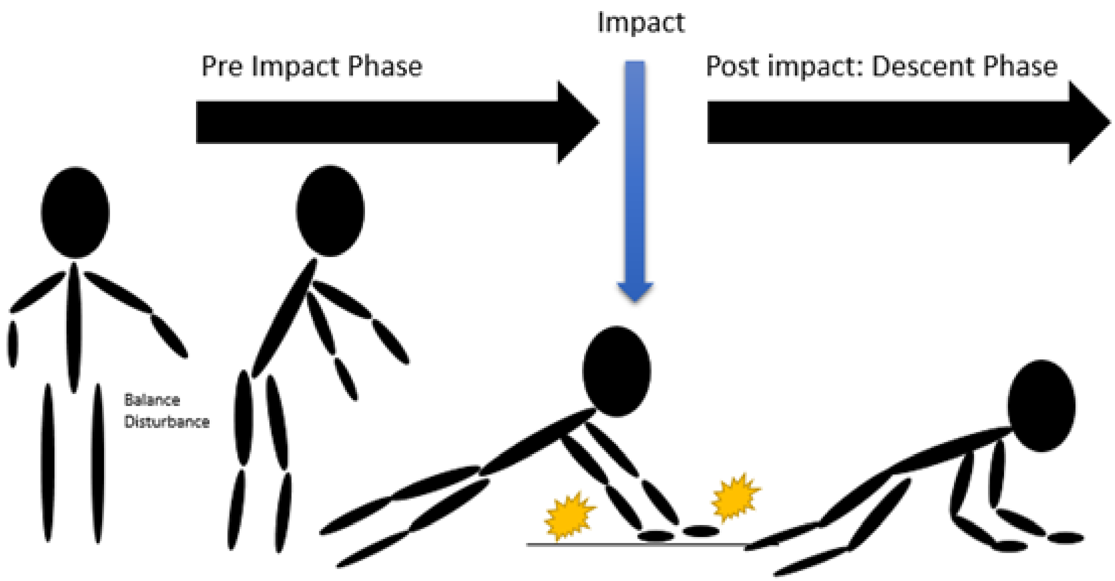


Figure 2.1. Diagram of forward fall arrest stages; Pre impact phase, Impact, Post impact

2.1.3.2 Fall arrest strategy

During a forward fall, loads on the hands can reach 2-3 times body weight from half standing height [22]. At impact, increased peak wrist impact forces are associated with greater shoulder flexion and less elbow flexion [56]. DeGoede et al [57] found that young men could significantly diminish impact forces on their outstretched hands by learning to control the acceleration of the upper extremity to the ground, and by slightly flexing the elbow prior to impact.

Chou et al.[40] found that if the elbow flexed upon impact, the time to peak force was delayed, resulting in greater impulse. Impulse is defined as the product of the force and the length of time that the force is applied and is the area under the time/ force curve. In a forward fall the musculoskeletal system can absorb more impulse by delaying the time of the second peak in force and decreasing the value of the first peak force.

The risk of injury in a FOOSH is related to body position and impact velocity [20]. Hand impact velocity is strongly associated with the peak force during simulated forward falls [22]. Risk of fractures in the UE is said to be influenced by impact velocity and elbow flexion [40,57]. Voluntarily increasing the elbow flexion angle by 40° and reducing the relative velocity between the hands while arresting an oncoming mass reduced the impact force from over 300N to less than 120 N [57]. From this study of a pendulum load hitting the hands tested in younger men, DeGoede et al.[57] calculated a 0.9 percent/ degree decrease in impact force for an increase in elbow flexion. Since movement of joints takes place by contraction of muscles around the joints, muscle activity that is causing the protective postures of the UE is important to consider.

2.1.4 Gender and age differences for forward fall related injury

Fall related injury rates are 40-60% higher, and fracture rates are more than twice as high, in women than men of a similar age [15]. When faced with a fall, women are three times more likely to fall on their hip [47]. Women between the age of 50 and 65 years are more likely to use their hands during a fall compared with women over 65 years [47]. Women over 65 years are half as likely to first use hand impact when falling compared to men [47]. Interestingly, women seem to use their hands less often for arresting falls than men yet, the rate of upper extremity fracture rates are far higher in women than men [47,58]. Women are more likely to fall forward [22,23] and are 4 – 6 times more likely than men to fracture their wrist; most commonly as a result of a forward fall [59]. In older women, the most common site of fall related fractures are the upper extremity, hip, and the trunk in that order [58]. Distal radius fractures are associated with significant impairment, functional limitations, non-unions, secondary osteoarthritis and chronic pain [60]. Wrist fractures are commonly the first fracture sustained by early post-menopausal women, and are a known risk factor for a future hip fracture [10].

2.2 Research evaluating forward falls in older adults

2.2.1 Video analysis of actual falls

Video analysis studies provide a unique opportunity to capture the circumstances and factors associated with actual falls. A Canadian research team has extensively employed video analysis to understand the frequency of falls and the factors that are associated with fall induced injury in long-term care [48,61-63]. The activity that most often preceded a fall was walking forward [62]. Landing on the hands, a protective mechanism to arrest the body and avoid injury to the head, has been observed in 74% of falls videotaped in long term care [48]. Of the falls observed in this study the odds of head impact were greatest for forward falls and hand impact was not associated with the reduction of head impact [48]. The authors inferred that, while the protective use of the hands is an automatic response occurring in most forward falls, this response was relatively ineffective at reducing head impact which occurred in 37% of the falls [48].

2.2.2 Lab based studies

Effective fall arrest involves three stages: Pre impact movement of the arms during decent into a position in preparation for hand impact; impact of the hands; and the post-impact descent and energy absorption in the upper limb joints (e.g., shoulder and elbow) to arrest the downward movement of the body (and prevent head impact). Previous studies measuring forward fall biomechanics and reaction time in men and women are summarized in Table 2.1. In a laboratory setting, a sample of young adults (male and female from 22 to 35 years) effectively used their hands on the contact surface with wrists extended prior to impact in a forward fall causing avoidance of impact of the head, pelvis or hip [50,51]. DeGoede et al [64] observed that when women were required to stop an approaching target with hands starting at thigh level, older women were 20% slower to react than younger women. The ability to react quickly seems to be of importance during forward falls in order to place the hands in an optimal position on the ground or other surface, However, quicker motion may not necessarily result in an improved descent as it has been observed that faster arm movement prior to landing in unexpected falls resulted in abrupt and rigid post-impact response with higher force peaks [41]. It may be that not only reaction time but the pre activation of the UE muscles and ability to develop appropriate torques may be more essential to decrease impact force and safely arrest the body. There seems to be differences between self-initiated falls and cable-released forward falls in young and older males in a laboratory setting. Self -initiated falls tend to have smaller force peaks and longer breaking times (softer landings) [41]. Cable release falls, which could be argued to be more related to actual fall situations, were characterized by higher force peaks and shorter breaking time (stiff landings) [41]. In the same study, older men exhibited higher force peaks and shorter breaking time than the young during the cable release falls. Older men also had less wrist extension and more elbow extension at impact. Also, the elbow and shoulder flexion velocities were much higher in older adults, resulting in a much more abrupt and stiff arm landing [41].

During a FOOSH an optimal goal would be to absorb enough energy with the UE to reduce the risk of impact to the head or torso. It has been observed that older women absorb 45% less energy during controlled descents on the hands with their bodies positioned at varying angles from vertical [29]. Older women absorbed the most energy with the trunk at a starting descent angle of 56° from vertical and the maximum angle achieved was 67° while the younger were able to achieve 87° [29]. Muscle activity and strength were not measured making it difficult to determine why older women had more difficulty absorbing the energy of descent [29].

After an intervention of teaching young male participants to minimize the force on their hands by "catching the floor" during the descent. DeGoede et al. [22] observed a 27% decrease in impact forces. This study concluded that participants could volitionally reduce the peak ground reaction force by flexing the elbows and reducing the velocity of the wrist relative to the trunk [22]. Another study by the same research group investigated an intervention of falling technique further. Other falling technique cues they used were "avoid accelerating your hand into the ground at impact" and "land with a slightly flexed elbow angle; do not ever land with a straight elbow angle". After a 10 min intervention, young males were able to reduce their hand impact force by 18% which is less than the original 27% they observed in the first study. The authors accounted for this difference by indicating that in the first study they permitted the participants to have torso impact following hand impact [22,42].

Past laboratory studies of forward falls have investigated fall strategies and heights on kinematics in different simulations where participants were aware of the fall. Most studies have observed male participants and not age related changes in women specifically.

Table 2.1 Summary of Laboratory studies investigating forward falls.

Study	Population Measured	Outcome Measures	Findings	Limitations
Chiu et al., 1998	Healthy men and women aged 20-35 (n=8) per group. Combination of experimental and mathematical models	Impact response for different fall heights(1,3,5cm) Shoulder stiffness, Wrist stiffness, Force peak one, force peak two, effective stiffness and dampening of the shoulder and torso.	Falls were characterized by a high freq. peak occurring 20ms after impact and a lower freq. peak occurring 110 ms after impact. As fall height increased, so did the first and second force peaks. Shoulder stiffness decreased with increasing fall height (because of elbow flexion).	Did not assess elbow stiffness. Release experiment was from knees and not unexpected.
Robinovitch et al., 2005	Young women (n=30; aged 18-35) elderly women (n=30; aged 70-88)	Standing forward and sideways with shoulder height targets. Measure reaction time (bilateral).	A typical elderly woman should be able to move her hands quickly enough to break a forward fall, but not a sideways fall. Older women are slower than young.	No muscle activity measured (onset times) Did not replicate a true fall scenario.
Chou et al., 2001	11 young males (aged 20-30)	Ground reaction force, joint forces, and joint moments. Fall heights (3cm, 6cm). Two different elbow postures. (Full extension and flexed	Elbow flexion at impact significantly reduces the first peak impact force value and postpones the maximum peak value.	Did not assess muscle activity. Did not include women. Did not

		immediately during the downward phase.		assess age differences. Sample only includes young males
DeGoede et al., 2002	Healthy young male (age 22-28) n=5	Fall simulations during a "natural fall', consisting of elbow flexion on impact, and "stiff arm fall". Measured first and second maximum force after impact, elbow angle, vertical velocity, and vertical velocity of distal forearm at hand contact. Measured triceps and biceps EMG Averaged over 50ms prior to hand contact.	Peak force correlated with elbow angle at impact, wrist velocity and pre-EMG triceps activity. Flexing the elbow and reducing velocity of the hands relative to the torso has the greatest ability to reduce peak forces.	Small sample size of young males.
Kim et al., 2003	Ten health older males (mean 66.4) and 10 young males (mean 24.1)	Different falling modes (self-initiated vs. cable-released; Two force peaks, joint angles and velocities.	Self-initiated falls demonstrated smaller force peaks, longer breaking peak times. Cable released demonstrated increased force peaks and shorter breaking time. During the cable released falls the older adults had less wrist extension and more elbow extension at touchdown. Older adults had significantly more elbow extension and shoulder flexion angular velocities. This resulted in harder impact and abrupt cessation of the body movement.	Male only population.
Lo et al., 2003	29 healthy young males (mean age:23)	Three-month intervention (learning to reduce hand impact forces).	Instructed to "reduce your elbow extension speed prior to hand impact Post intervention, young males were able to reduce impact by 18%. Force can also be reduced with a 10-minute instruction.	No information on muscle activity changes with interventions. Male only data.
Burkhart et al., 2013	Fall simulation; 10 males (mean 22.8) and 10 females (mean 25.6)	Different fall types; straight arm; instructed to minimize impact (self-selected); bent elbow (20° flexion). Measured peak impact force, impulse, impulse duration, peak accelerations, and peak acceleration rates. Muscle activity of the biceps brachii, brachioradialis, triceps brachii, anconeus, extensor carpi ulnaris, flexor carpi ulnaris during the start of data collection to release; from release to impact; impact to peak force' peak force to end of impulse.	Mean force was significantly less for 0.05m fall heights compared to but peak force was unaffected. Straight arm falls produced significantly greater impulses and load rates and shorter impulse durations than self-selected and bent arm falls. EMG was significantly different between all fall phases. All muscles exhibited pre impact preparatory muscle activity. Anconeus muscle activity was significantly greater during bent arm falls compare to the other types. Extensor carpi ulnaris muscle activity was significantly greater during straight arm falls compared to the self-selected falls.	No measurement of proximal muscle activity. Did not include women over 65

2.2.3 Computer simulation models to predict neuromuscular capacity for successful landing

To avoid the level of risk associated with laboratory studies of falls in older adults, validated biomechanical models have been used to estimate the kinematics, muscle strength requirements and peak impact forces at the forearms. An early mathematical model that employed a fully extended elbow during impact found that peak force was entirely determined by impact velocity and the dampening properties of the hand and ground [52,65]. Chiu et al.[52] calculated energy absorption, defined as the area under the force-displacement curve at the wrist and shoulder. They found that peak energy absorption and displacements were greatest at the shoulder, but the model was limited because it did not include the elbow joint. Computer simulation studies have suggested that age related reductions in elbow extensor strength reduces the ability of older women to use their UE to arrest a fall during the descent phase [23]. Using validated computer simulations in the healthy younger adults, wrist velocity prior to impact and elbow angle at impact have the greatest influence on ground reaction force in the models [23].The peak force acting at the wrist was largest for a landing with a more extended elbow and higher impact velocities [23]. DeGoede et al. [23] indicated that inadequate arm extensor strength can lead to a slightly flexed elbow to buckle under the force at hand impact. When the elbow buckles, this could result in torso or head impact. Model predictions indicated that mild velocity falls (0.5m/s) in older women could not be arrested without a torso impact [23]. Further research is needed to understand the arm strength and energy absorption demands of a fall descent.

2.3 Factors associated with forward fall dynamics and injury risk

2.3.1 Fall risk factors

Forward falls are the result of the interactions of several factors, including environmental, physiological and situational factors. Environmental factors such as curbs and obstacles are important contributors to forward falls as they may present tripping hazards while walking forward. Weather conditions such as ice, rain and snow are well known to provide increased opportunity to fall [66,67]. Physiological fall risk factors such as health conditions, polypharmacy and impairments such as balance, gait and muscle strength also play a role in fall risk [68]. The female gender, advancing age, gait and balance deficits, chronic disease, and medication use, have been associated with a higher risk of falling [14]. Two primary intrinsic impairments associated with increased fall risk include balance and muscle strength.

2.3.1.1 Balance and falls

The ability to maintain balance involves the interaction of the neuromuscular, proprioceptive, vestibular and visual systems. Age related declines in these systems increase risk of falls in older adults [69]. Balance is a strong predictor of fall risk and the ability to control the trunk muscles is key to balance, particularly the ability to avoid a fall when rapid compensatory responses are needed after a slip or trip [70]. Of particular interest to this thesis, is the contribution of these rapid compensatory responses, known as postural control to fall dynamics and impact forces that may influence the injury severity when a fall cannot be avoided.

Postural control can be defined as a process that involves rapid, automatic integration of information from the neuromuscular, proprioceptive, vestibular and visual systems in the presence of cognition which includes attention and reaction time [71]. Postural control consists of both anticipatory and reactive balance strategies. Anticipatory balance is defined as muscle activity in postural muscles prior to postural perturbations to counteract the expected mechanical effect in a feedforward manner [72,73]. Reactive balance involves the activation of postural muscles after an external disturbance to ensure balance recovery [74]. The initial neuromuscular state prior to the initiation of the fall has been found to have a significant impact on reactive balance and the ability to attenuate forces in the extremity and, thus, to the risk of injury. Reactive balance is of particular interest for this thesis, as an UE protective response is a key part of a FOOSH and resultant injury. A better understanding of the modifiable factors associated with this protective response could help to better identify the factors that could lead to avoidance of injury.

Reactive movements of the trunk in combination with UE protective strategies, have been shown to decrease the risk of impact on the pelvis during a fall in young adults [50]. This alludes to the suggestion that effective movement and muscle activity strategies exist for preventing injury during a fall. Given the complex interaction of the body's muscles in controlling motion, a combination of both trunk stabilization and upper limb activation would theoretically improve the ability to lower the body in a controlled manner to avoid injury. The UE neuromuscular control in both of these strategies is reviewed later in section 2.3.3.1.

2.3.2 Muscle morphology changes with aging

With aging, body composition changes towards a fat mass increase and lean body mass decrease [25]. These changes are observed even when body weight and physical activity remain constant [25]. The prevalence of sarcopenia, defined as a combination of both low muscle mass and low muscle function [75], has been observed to range from 10-40 % loss of lean tissue in older people and the prevalence seems to depend on age and sex. Muscle strength regressions with age appear to be a result of declines in muscle quality which may be attributable to age-related changes in neurologic and skeletal muscle factors such as loss of motor neurons, decreased neuromuscular transmission, muscle fibre morphology, and adipose tissue infiltration between muscles and muscle fascicles (inter- and intra-muscular adipose tissue) [25].

2.3.2.1 Sarcopenia and dynapenia and falls

As skeletal muscle mass declines with aging, muscle strength also decreases [76]. Some studies have observed a weak association between reductions in muscle strength and muscle mass in older adults [76,77]. In both the lower and upper extremities, Beliaeff et.al. [78] concluded that declining muscle mass is not the major contributing factor to the age related loss in muscle strength in older adults. The relationship is nonlinear so that the strength decline exceeds the decline in muscle mass by a factor of 3:1, and muscle mass no longer closely reflects muscle strength [76]. There is debate that the age-related loss in muscle strength should be described independently as a new condition called dynapenia. This distinction is based on the evidence that the loss of muscle mass and strength are two distinct processes with different pathophysiology [79]. Some observational studies have consistently shown a strong association between muscle strength and function: including falls and mortality, with poor associations between muscle mass and function measures [80,81]. There seems to be a paradigm shift toward seeing muscle weakness as a major risk factor for functional limitations and fall status in older adults as opposed to muscle mass. Additionally, other components of muscle performance such as power and eccentric strength may be more directly associated to fall risk (refer to Section 2.3.2.2). Narici et al.[82] indicated that the reduction in single fibre specific tension is one of the major factors contributing to the decline in intrinsic muscle force. Recent evidence suggests that this could be the result of decrease in the number of actin myosin cross-bridges and a reduction in excitation-contraction coupling [83]. Another suggestion for the decrease in muscle force (strength) is a decrease in neural drive to the agonist muscle and an increase of neural drive to the antagonist muscles [82].

Some studies have found no relation between muscle mass and functional status and mortality whereas muscle strength is an independent predictor of function including falls and mortality [80,84-86]. Clark and Manini [79] developed an algorithm to help define dynapenia that includes measurements of hand grip and knee extension strength [87].

The variation in the method of measurement as well as the muscle group measured makes it difficult to draw conclusions. Handgrip and lower limb strength have been the primary sites measured [88-91]; both identified as the most consistent predictors of falls and fractures in both sexes.

2.3.2.2 Muscle contraction type and aging

Skeletal muscles produce tension through static and dynamic contractions. Concentric contractions involve the shortening of the muscle during contraction and conversely eccentric contractions involve lengthening of the muscle during contraction, while isometric involves no change in length [92]. These contraction types are affected differently by aging. This was first observed in older women who had maintained their eccentric knee extensor strength to a greater degree than their concentric strength when compared to young women [93]. Older adult’s strength during concentric and isometric contractions has been observed to be 50% less than younger adults, while eccentric deficits in strength have been observed at only 20% in men and absent in women [94-96]. At the whole muscle level, force loss can be attributed to shortened fascicles, less pennate fascicle organization and less tendon compliance [93]. The decrease in isometric strength can be attributed to cellular and whole muscle changes [93]. There has been evidence to suggest that the decreased in concentric strength in older adults is the result of greater activation of antagonist muscles [97]. Differences in eccentric strength in older adults compared to younger adults has been reported between 2-50% and average difference of 20 % [92]. Maintenance of eccentric strength in older adults can be attributed to both neural and mechanically mediated mechanisms [92]. Hortobagyi et al [94] suggested the reason for eccentric preservation was due to the greater amount intramuscular connective tissue observed in older adults which may provide increased passive resistance during muscle lengthening. This increase in passive resistance (stiffness) could explain why concentric and isometric force production is not affected. Eccentric preservation is also consistent across a range of angular velocities as older adults have a velocity dependence of absolute eccentric strength [98]. Power et al [98] found no age related differences in eccentric strength at velocities from 45°/s - 360°/s.

Muscle power reflects the ability to produce force quickly and slowing of contractile properties with age can slow the rate of force development. Maximal voluntary muscle power also declines with aging [99]. Improvements in leg power have been observed to produce better outcomes in composite functional measures (Short Physical Performance Battery) when compare to leg strength [100,101]. Power declines have been attributed to impaired excitation-contraction properties [102] and slower muscle contractile coupling [103].

Other factors such as neuromuscular activation are also involved in the age related reductions in muscle strength and power [104]. Neuromuscular activation can modify muscle force producing capabilities by altering the number of active motor units firing during muscle contraction. For example, co-activation of agonist and antagonist muscles has been observed to occur during some dynamic functional tasks, potentially a mechanism to increase joint stabilization [105,106]. It has previously been shown that older adults exhibit increased co-activation of knee muscles during walking and that this may contribute to increased metabolic costs for locomotion [107]. Older adults employ co-activation patterns in the muscles about the ankle as a compensation to increase joint stiffness and to maintain balance during walking [108]. A delay in muscle activation rate has been linked to impaired capability of older adults to produce force rapidly under isometric conditions [109-111]. Maximum voluntary muscle power declines with age and is predictive of functionality in older adults [99,112]. Clark et al. [113] confirmed that the rate of neuromuscular activation is associated with dynamic muscle performance and mobility function in older adults. The slowing of muscle activation rates can be increased with resistance training [114] and activation rates are higher in older adults who maintain a physically active lifestyle [115].

2.3.2.3 Regional differences in aging changes of muscle strength

Age-related declines in muscle strength are not uniform across muscle groups. Muscle mass declines in the lower limbs are most prominent with a decrease of 30-40% between the second and eighth decade of life [116,117] with greater lower extremity strength declines in knee extensors and flexors in women compared to men [118]. Upper body strength, such as for elbow flexors and extensors seems to be better maintained with age in men compared to women [104,119,120]. It has also been observed that hand grip strength is better maintained with age compared to knee extensor strength [121,122]; however, few studies have investigated either age or sex related differences in muscle strength decline in the UE, particularly at the elbow or shoulder.

2.3.2.4 Quantifying muscle performance in older adults

Muscle strength quantification in older adults has been primarily reliant on dynamometric measures of hand grip and knee extension strength [123-131]. Handgrip strength is commonly used clinically as an estimate of generalized muscle strength because of cost effectiveness, portability, simplicity and time-efficiency [132]. Handgrip strength has been associated with increased risk of physical disability [123], osteoporosis [128], poor mobility [124], cognitive decline [127] and mortality [133]. Historically there has been limited understanding of the relationship between handgrip strength and muscle size in the forearm and upper arm [134], a recent study found forearm muscle thickness as measured with ultrasound was correlated to handgrip strength and knee extension strength in older women [126].

Martien et al.[132] demonstrated that handgrip and knee extension strength are important predictors of functional performance in older adults with knee extensor strength a better predictor of functional decline in assisted living as compared to community dwelling older adults or those residing in nursing homes. Handgrip strength has been the focus of many studies on functional and strength in the upper limb of older adults [121,135-138]. One study found that handgrip strength in women older than 75 years has been moderately correlated with overall strength but weakly correlated with the timed up and go test (TUG) [136] suggesting limited ability to predict walking performance. This suggests that handgrip strength does not fully represent functional ability in older adults, but forearm strength is a good potential marker for functional decline and fall risk status.

Age-related limitations of activities of daily living (ADLs) may result from several factors, including reduced muscle strength and decreased muscle mass [117]. As well, slowed muscle activation may particularly affect functional tasks require fast adjustments to muscle activation; this may pose a particular challenge when task demands are higher such as during fast walking, stair negotiation, and balance recovery after tripping [19].

Functional fitness can be defined as the capacity to perform activities of daily living safely and independently without fatigue or pain [139]. Each individual's ability to perform ADLs can be tested using a range of objective and self-reported outcomes. In the upper limbs, age-related decline in the ability to carry objects with the arms seemed to be primarily a function of muscle size and strength [140]. Arm activities that require a strength component are important for performing ADLs in older adults such as getting up from a chair or lower surface, maintaining balance after a perturbation, reaching and fine motor activities; yet few isokinetic UE strength studies in older adults exist [137,141]. Studies in men have reported that isokinetic testing of the elbow flexors was found to be more sensitive to age-related changes compared to isometric and isotonic [141]. One study found the most highly correlated strength measure to performance of daily activities in the upper extremity was shoulder external rotation strength while grip strength was not significantly associated [142]. There are no studies that have compared other muscle groups such as the shoulder girdle and upper arm and the relationship to fall injury risk status. Attention needs to be given to functionally significant measurements of strength in the upper extremity in regards to fall injury risk.

Dynamic strength and power assessment of older women are limited because of safety and logistical concerns, and have usually focused on the lower extremity [122,125]. Muscle power is defined as the product of force and velocity of movement and plays a fundamental role in lower extremity functional performance [143]. Isokinetic dynamometry is the gold standard for measuring muscle strength in the literature but poses feasibility difficulties as a clinical tool to measure muscle performance of older adults due to the cost and space constraints of the equipment [144]. Additionally, isokinetic dynamometry strength assessment in the UE is commonly restricted to single joint movements with a fixed axis of rotation with a lack of dynamic multi joints movements and dynamic tasks associated with activities of daily living. There is a gap in the literature concerning multi-joint upper extremity functional strength tests that represent functionally relevant movements. Age-related declines in strength and power in specific, functionally related contraction types also need to be studied further.

In summary, despite gold standard equipment available to accurately assess UE strength, measurement tools lack face validity to assess functional strength requirements of the upper limb associated with the demands required to arrest the energy of a forward fall. Possibly what is even more important to add to the evaluation of functional muscle performance is the measurement of neuromuscular activation during the functional task.

2.3.3 Neuromuscular activation

Neuromuscular activation can be defined as the process by which the nervous system produces muscular force through recruitment and rate coding of motor units [4]. A common method used to quantify muscle activity during functional tasks is electromyography (EMG). EMG detects the bioelectrical activity associated with muscle contraction. The least invasive form is surface EMG which assesses timing and magnitude of activation. EMG can provide information about muscle recruitment patterns, about the relative timing of muscle activation and about muscle activation levels relative to a voluntary isometric maximum or another task. It may be that neurally mediated changes in muscle excitation patterns have functional implications such as controlling the downward motion of fall descent. The advantage to this method of measurement is the ability to visualize muscle activity of several functionally related muscles during a specific task of interest. The disadvantages are that while evidence suggests a link between neuromuscular activation rate and static force production [145], influence of neuromuscular activation and muscle power is not well understood**.** One longitudinal study did find that age-related reduction of EMG rise contributes to the loss of muscle power [143].

Disadvantages are that EMG activation levels are not linearly related to either force generation or power. There is a positive relationship between isometric muscle force and surface EMG amplitude [146,147]; however, EMG does not yield mechanical properties of muscle activation such as muscle length changes, connective tissue resistance and fatigue which affect force generation [4]. Motor unit recruitment and firing frequency are both reduced in older adults [82,148]. The reduction in muscle fibre size has a preferential reduction to type II fibres in older adults.

Older subjects exhibited 10-40% smaller type II fibres than young controls [149]. The changes to type II muscle fibre in the older adults can be explained by the remodeling of motor units that result from the denervation of type II and re-innervation of type I [150,151]. In summary, aging can lead to declines in neuromuscular function during both dynamic and static force productions [152].

2.3.3.1 Neuromuscular activity during the phases of fall descent

While the UE plays a key role in impact absorption, the control of the landing involves co-ordination of the whole body. The three phases of a fall discussed in 2.1.3.1 with corresponding muscle activity of fall descent includes: **Pre impact**: 1) anticipatory postural response to the unexpected perturbation of the fall, 2) movement of the arms during descent into a position in preparation for hand impact (reactive postural control), 3) pre impact neuromuscular state of the limbs and trunk; **Impact**: 4) impact of the hands including the immediate post impact muscle activity for energy absorption in the upper limb joints (i.e., shoulder and elbow) and **Post impact:** 5) descent to arrest the downward motion of the body (and prevent head impact) and to lower the body to the floor or surface.

Phase 1: Pre impact The aging process is associated with a general decline in neuromuscular functions such as slowing of axonal conduction velocity which can impair speed of motion resulting in diminished balance control [93]. To preserve the equilibrium, the central nervous system (CNS) uses two types of alterations in the activity of trunk and leg muscles; feed forward and feedback mechanisms. Anticipatory postural adjustment (APAs) is a well-known phenomenon expressed as the change in the activity of postural muscles approximately 100 ms before the posture is disturbed [153]. APAs are associated with the activation or inhibition of trunk and leg muscles prior to the actual perturbation of balance [154]. APAs minimize the negative consequences of a predicted postural perturbation [73]. Feedback strategies include muscle activity evoked by postural perturbation.

Older adults do not adapt and use anticipatory postural mechanisms to the same extent as younger adults in response to sudden and unpredictable or externally-triggered perturbations in the anterior/posterior direction [155]. It has been observed previously that feedback postural responses can delay or contribute to voluntary movement [156,157]. When postural perturbations are associated with forward voluntary reach movements, the initial postural response is maintained, which [158] suggested that the nervous system prioritizes maintaining a stable postural base. It is said that anticipatory adjustments can create rigidity of the body segments not directly involved in movement and this can facilitate subsequent movement which will be addressed in phase 2 [159]. Muscle activity in preparation for impact appears to be a strategy to prepare the muscles to absorb the impact [54]. In the lower extremity the goal of such pre-landing activation was said to be to adjust muscle stiffness prior to landing, in order to minimize the impact force on landing. In the lower extremity landings after unexpected falls result in larger ground reaction force (GRF) then landing after expected falls, a result that seems to correlate with differences in pre-landing muscle activity patterns [54,160]. Animal studies have led us to believe these pre-landing muscle contractions were thought to be pre-programmed rather than reflex in nature, and under the influence of the supraspinal centers for the control of the early deceleration phase associated with landing [161]. To prevent a slightly flexed elbow joint from buckling under impulsive load at the hand, arm muscles need to be pre-contracted. Pre contraction is deemed necessary because the ground reaction force on the hand peaks too rapidly [20,22]. During a forward descent on outstretch arms the triceps brachii is undergoing a lengthening contraction which assists to control the amount of elbow angle that is achieved at impact. Pre impact neuromuscular activity of the triceps could be important to investigate during a forward fall.

Phase 2: Impact Rapid arm movements are important components of the natural defense against a fall. Impairments of initiation of hand movement into a protective position could decrease the ability to arrest a forward fall. Not many studies have evaluated age related changes in postural arm reactions. One study found older women 20% slower than younger women when moving hands from thigh level to arrest an oncoming object [64]. Similar results were reported by Robinovitch et al.[162] citing the deficiencies were observed in movement time (time to execute motor response) rather than reaction time (time to initiate at response) with 83% of the older women still meeting the arm movement time requirements to break a fall. McIlroy et al [163] observed that arm muscle responses can in fact precede those of leg muscles and effectively help prevent a fall or cushion the impact. Older adults are more likely to initiate arm movement and grasp safety rails for support but are less able to execute a reach rapidly [164]. Reaction time studies have shown that healthy older women are slower than young women at initiating hand movement but are still able to move hands in an appropriate position [162] but little is known about the required muscle activity required to arrest the body and control descent in order to avoid impact of the head or torso.

Phase 3: Post impact Impact of the hands and post impact muscle activity contributes to the later stages of breaking the fall and stabilization of body posture [54]. Aging results in an increased instantaneous stiffness (reduction in elasticity) in whole muscle as well as in single fibers [150,165]. This may be due to an increase in the number of cross-bridges in the weak-binding state but other factors such as changes in cross-bridge compliance and sarcomeric elements like titin may contribute by increasing stiffness [150]. Increased antagonist co-activation will decrease overall net joint torque whereas muscle pre activity during a lengthening contraction is said to increase torque [166]. The triceps brachii muscle group are responsible for the stiffness of the elbow when forcibly flexed as would occur after impact [167]. Muscle forces are an important force attenuating component, which can actively adjust the amount of shock attenuation through eccentric contractions about the joint. In both expected and unexpected impacts the central nervous system (CNS) must modulate muscle force before contact with landing surface as well as throughout the time course of force application [54]. Stiffness of a joint is generated by active (muscle contractions) and passive mechanisms (visco-elasticity of muscle and tendon stiffness) [168]. Muscle stiffness contributes to joint stability through mechanical and neural aspects and increased muscle stiffness is said to increase performance and decrease the risk of injury [169,170]. Stiffness of a joint in the lower limbs is said to provide torque that restricts the perturbation effects and thereby reduces the probability of a fall [171]. It has been advocated that some level of limb stiffness is beneficial to enhance athletic performance, however too much or too little stiffness may increase the risk of injury [172]. Elbow joint stiffness does vary with sex, ability to generate triceps co-contraction and the elbow angle at impact when near extreme extension [20]. Given the complex interaction of the body's muscles in controlling motion, a combination of both trunk stabilization and upper limb activation would theoretically improve the ability to lower the body in a controlled manner to avoid injury.

2.4 Trunk stability and fall injury risk

In addition to UE muscle activity, trunk muscle activity could contribute to the control of the descent. A descent on outstretched hands is a closed kinetic chain task which places load on the UE but also on the thoracic and lumbar spine [173]. The post impact descent phase is similar to the eccentric lowering phase of a "push-up" and high levels of activation of the oblique abdominal muscles has been observed during push-ups in young men. [173,174]. Contraction of the trunk muscles occurs in both preparation for and in response for spinal loading [175]. In a forward fall it would be expected that trunk muscles would activate in anticipation and in response to the impact. Contributions of the trunk musculature has not been previously investigated during a controlled or unexpected descent in women. Trunk activation may be an important component to control body descent during each of the 5 phases of a FOOSH as discussed in section 2.1.3.1. Hodges et al [176] observed that the onset of transverse abdominis (TrA) EMG preceded the onset of anterior deltoid in only the fast movement condition when compared with a slow and self-paced arm flexion. Impairments in the initiation of UE movement time in the older adults could be indicative of delayed onset latencies of the trunk musculature and could be important to investigate during the pre impact phase. During the impact and post impact phase the trunk muscles are likely contributing to the stability of the spine and aid in controlling a safe descent phase.

Trunk muscle activation could be important to investigate during a FOOSH because trunk muscle activity creates stability of the spine and therefore enables proximal to distal force generation that is said to protect distal joints. Proximal muscle activation has been also seen to produce increased levels of muscle activity distally. A comparison of trunk muscle activation in young and older women during a simulated fall, with an unexpected release, could emphasize age-related differences in fall response. These differences might influence the risk of injury and the findings could be used to inform injury-prevention strategies.

2.4.1 Trunk muscle performance and relationship to stability

Trunk stability is essential for the development, transfer and dissipation of forces throughout the kinetic chain. Stability of the trunk is necessary to maintain the integrity of the spinal cord, provide resistance to perturbation and provide a stable base for movement of the extremities [175]. The importance of this proximal musculature is the also transfer and dissipation of forces occurs from the trunk to the extremities and from the extremities to the trunk. Most research has been focused on the importance of core and trunk muscle strength as it related to performance during sport-related and everyday activities. Researchers have established a general understanding that functionally the trunk is a kinetic link that facilitates the transfer of torques and angular momentum between upper and lower extremities during the execution of whole-body movements such as sports skills, occupational skills, fitness activities, and activities of daily living [177]. Kibler et al. [178] stated that trunk strength is especially important in most daily activities, including sport and recreational activities, because it provides stability for distal mobility. The transfer of forces from proximal to distal or distal to proximal requires adequate muscular capacity (strength and endurance) and central nervous system programming that produces synchronous activation of the muscles [178]. Some studies suggest that increased trunk stiffness may be beneficial in maintaining balance [171]. Other studies indicate that increased trunk stiffness may hamper performance of compensatory movements after a mechanical perturbation and compromise balance control [179]. Van der Burg et al.[171] concluded that an artificial trunk stiffening corset appeared to have no net effect on balance recovery after a trip in young adults. Mobility of the spine is also important to compensate for, and attenuate disturbances of posture. When control of the trunk muscles is altered, spinal mobility and the contribution of the trunk muscles to posture may be impaired and balance can be compromised [180]. Hodges et al believe that anticipatory postural adjustments involve movements and not rigidification of the trunk [181].

2.4.2 Trunk stability mechanisms

Early research in core stability performed by Panjabi [182] and Bergmark [183] labeled three subsystems that create spinal stability: the passive osseo-ligamentous system, the active muscular system and the neural control system. The passive system is comprised of the boney and ligamentous structures, with their abundant mechanoreceptors, that relay information concerning spine position and movements from the facet joint capsules and ligaments of the vertebral column to the CNS. By itself, this passive system has been shown to "buckle" or "fail" in vitro at loads of less than 100 N (approx. 20lbs) [184]. The active muscular system includes the muscles and tendons that generate the forces that create spinal stability. This system contributes to stability by developing intra-abdominal pressure, spinal compressive forces, and hip and trunk stiffness [175]. The trunk musculature has been described as a box; with the abdominals in the front, paraspinal and gluteal muscles in the back, the diaphragm as the top, and the pelvic floor and hip girdle muscles as the bottom [185].The core serves as a muscular corset that works as a unit to stabilize the body and spine, with or without limb movement. The trunk muscles can be classified into two groups: global stabilizers (larger muscles with longer moment arms, such as internal/ external oblique and rectus abdomius) and local stabilizers (smaller intersegmental and postural control muscles such as multifidus, transverse abdominus (TrA) and the pelvic floor muscles). The global muscles produce torque and transfer loads between the thoracic cage and the pelvis; while the local muscles are associated with the segmental stability of the spine and postural control [186-189]. Co-activation of both the global and local muscles contributes to trunk stability [190].

The TrA is a key spinal stabilizer; EMG has demonstrated that it is active during all trunk motions [191]. The TrA is the deepest of the abdominal muscles, running transversely around the abdomen from the linea alba to the transverse processes of the lumbar vertebrae via the thoracolumbar fascia. The main roles of this muscle are to increase tension on the thoracolumbar fascia [192] and to generate intra-abdominal pressure [193]. Intra-abdominal pressure decreases the perturbation of trunk segments caused by a sudden load and it increases as the load increases [194-196]. Both TrA, internal obliques (IO), and external obliques (EO) share attachment sites such as the costal cartilages, lumbar spine via the thoracolumbar fascia and the iliac crest and pubis. Cadaveric studies have indicated that the IO and TrA are fused prior to inserting into the linea alba [197]. The magnitude of the muscle forces developed is monitored by golgi tendon organs and muscle spindles, which link the muscular system to the neural control system.

The neural control system resides in the CNS where it controls trunk movement and stability through both feedback and feed-forward mechanisms using information it receives from the proprioceptors in the active and passive systems. Feed-forward mechanisms are pre-planned motor programs in preparation for movement. Feed-forward is anticipatory in nature as the CNS activates muscles in preparation for an impending spine movement or load. Feedback to the CNS that is generated by elongation of a ligament or joint capsule and muscle tension development mediates adjustment from the CNS in muscle activation patterns to maintain spinal stability. Feedback mechanisms are used to fine tune motor programs, allowing skills to be performed with greater efficiency over time. The ability to quickly modulate the timing and recruitment of muscles in response to postural perturbations is considered to be paramount for maintaining balance and posture. Bugnariu et al. [155] suggested that feedforward postural adjustments of the abdominal muscles are likely organized at a low level of the nervous system and are initiated with limited processing time. Hodges et al. [181,193,198] found that feed-forward recruitment of the TrA preceded activation of all other muscles during movement of the upper and lower extremities with TrA EMG preceding the onset of preparatory motion of the trunk in all directions [181]. EMG studies have indicated that when exposed to unpredictable perturbations, the sequence of compensatory activation of the muscles was proximal to distal [154,199], which suggests that alterations in this pattern could result in a loss of postural stability. The ability to quickly modulate trunk muscle timing and recruitment of muscles in response to unexpected or expected postural perturbation is considered paramount in maintaining balance and posture. During self-initiated movements APAs are significantly delayed in the healthy older adults [200], with the postural muscles being recruited simultaneously with or after the activation of the prime mover [201].

2.4.3 Trunk stability and aging

Recent literature has suggested that the trunk muscles play an important role in balance and functional mobility in older adults. In older adults, decreases in trunk muscle strength and endurance have been found to be associated with poor balance and mobility [202-204]. Trunk strengthening interventions in older adults have produced improvement in dynamic balance [205] [202]. Specifically, after a 16-week intervention, improvements in trunk endurance, but not lower extremity muscle strength or power, were associated with clinically meaningful improvements in balance [202]. Trunk muscle atrophy may cause decreased trunk stability, resulting in increased fall risk and ADL disturbances [206]. Trunk muscle quality, assessed through fat infiltration, predicts functional capacity loss as well as the development or worsening of lower back pain, even when controlled for limb muscle size [207]. Rankin et al. [208] observed that there was a significant negative correlation between age and thickness of abdominal muscle in healthy subjects aged 20-72 years. A similar association was observed when independent older adults (independent in ADLs and walking) were compared to dependent older adults (chronically bedridden); EO, TrA, thoracic erector spinae and lumbar multifidus muscles were significantly thicker in the independent older adults group compared with the dependent older adults group [209]. The magnitude of decline for the TrA was greater in the dependent older adults at 51.5 % decline compared to the young group while the independent only had an 11.8 % decline. Since the TrA has an important role in stabilizing the lumbar spine, an increase in muscle thickness could have resulted in enhanced maintenance of the sitting position, as this study found that the dependent older adults group could not maintain this position. Muscle thickness of the rectus abdominus, external oblique, internal oblique, and transverse abdominus was examined across five age groups ranging from 20-85 year old women. Compared with young women, this study observed significant differences in the superficial muscles (Rectus abdominus, internal/external oblique) from the age of 25 (rectus abdominus) and from the age of 45 (internal/external oblique) [210]. These studies highlight age related abdominal muscle loss does occur in women, with possibly greater loss attributed to inactivity. Trunk muscle strength has also been related to instability and fall risk; however, fewer studies have examined this association. There are currently no studies that address how trunk muscle strength contributes to the ability to arrest a forward fall.

2.4.4 Electromyography of trunk muscles

Surface and fine-wire electrodes are two types of electrodes used to measure muscle activation [211,212]. Fine-wire electrodes sample a smaller number of motor units compared to surface electrodes. Fine-wire electrodes are recommended to be used on small or deep muscles to reduce cross-talk of surrounding muscles. Previous research regarding TrA activity has primarily involved the use of fine-wire EMG [181,191,193,198]. Surface electrodes positioned over the internal oblique (IO) muscle and TrA, located inferior to the anterior superior iliac spine have been demonstrated to represent the fine wire activity of TrA to within 10–15% of the contraction amplitude.[213] This is said to be the result of the combined activity of the TrA and IO as well as cross-talk from the rectus abdominus [213]. A more recent study with male participants validated surface EMG to fine-wire recordings for the feed-forward activation of the TrA/IO prior to rapid limb movement [201]. This study concluded that a separate and distinct signal or the combined activity of TrA/IO can be detected without the involvement to cross-talk from the rectus abdominus [201].

SUMMARY OF LITERATURE

The consequences of falls are diverse and both the physical and psychological aspects can greatly affect an individual's wellbeing [214]. Older women who fall and sustain injuries have considerably lower levels of wellbeing and are subsequently at a much greater risk of a future fall [214]. It is imperative to investigate modifiable fall risk factors in older women in order to both prevent falls and reduce injuries from unavoidable falls. Every year one third of adults over the age of 65 years will fall, of which 60% are forward FOOSHs [47]. A fall is one of the potentially serious consequences of sarcopenia [215]. In addition to muscle mass and contractile ability, age related changes in neuromuscular activation could be a critical component in the age-related differences in women. Arresting a forward fall with the upper extremities is demanding in terms of speed, co-ordination, motor control and hand placement. During a forward fall the impact force might be modulated by factors such as response time, upper extremity placement and neuromuscular capacity [21-23].There has been little research examining the age-related differences in neuromuscular and kinematic performance during a simulated forward fall. Considering all the documented age related changes in muscle performance and the high rate of injurious falls occurring in women over the age of 65 it becomes apparent that specific neuromuscular factors need to be studied during a fall. Therefore: the purpose of this thesis is to examine the age related differences in biomechanics, strength and muscle activity during a controlled and unexpected forward fall on the outstretched hands in women.

Biomechanical fall simulations have gathered information on specific fall strategies and variables that could lead to increased risk of injury during a forward fall in younger adults and men (Table 2.1). Similar simulated falling studies for older women have not been conducted; however, Sran et al. [29] found older women (average age 78 years) had significantly reduced ability to absorb the energy of a controlled body descent on the outstretched arm as compared to younger women. Given that DeGoede & Ashton-Miller [23] speculated that the age related decline in muscle strength substantially reduced the ability to control the impact of a forward fall, functionally relevant measurements of strength and power during contraction types could highlight deficits in the UE that could predict injury risk. Specifically, correlating functional strength deficits to biomechanical performance during a forward fall on outstretched hands could guide exercise prescription for injury preventions programs in women. Given that women over the age of 65 years are the population most often injured during a forward fall, there are very few studies evaluating the biomechanical and physiological factors associated with forward landing and descent in this population. The neuromuscular contribution of the upper extremity and trunk has not been examined during a simulated fall in an older adult population. Little is known about the neuromuscular requirements during pre impact and post-impact phases of the fall. Specifically, muscle activity requirements to safely arrest the forward momentum of the trunk have not been established in women. In addition, evidence in the age related changes in neuromuscular control during a FOOSH in women is not well known. A logical approach to this problem would first evaluate both muscle performance and the ability to absorb the impact sustained through the upper extremity when lowering the body during an expected controlled descent. The next step would be to test an unexpected descent since it has been tested in men but not women (Table 2.1). The purpose of previously identified studies described in the following three chapters were designed to discourse the gaps identified in the literature.

CHAPTER 3: STUDY 1

FEMALE AGE RELATED DIFFERENCES IN BIOMECHANICS AND MUSCLE ACTIVITY DURING DESCENTS ON THE OUTSTRETCHED ARMS

3.1 Introduction

In Canada, approximately one third of adults over the age of 65 years will experience at least one fall every year, and the total annual cost of fall-related injuries is approximately $6.7 billion [7]. Traumatic brain injury, due to impact to the head during the fall, accounts for 32% of fall-related hospital admissions and more than 50% of fall-related deaths in older adults [9,10]. Landing responses such as protective responses of the extremities may help to decrease the risk of head impact and injury. During forward falls, a common protective response is to arrest the fall (or stop downward movement of the trunk and head) through impact of the outstretched hands with the ground [20,48,50]. A range of 42%-60% of falls among community-dwelling older adults have been reported to be forward falls onto the outstretched hands (FOOSHs) [47,216]. We recently reported results from an analysis of 227 video-captured falls experienced by 130 older adults in long term care where 37% of falls involved head impact and the odds of head impact was greatest with forward directed falls. Hand impact was observed in 74% of falls, but was not associated with the reduction of head impact [48,63]. This signals the persistence with age in the generation and execution of upper-limb protective responses, but loss in the effectiveness of the response in older adults in long-term care. In laboratory simulated falls, head and hip impact is uncommon in young adults due to effective use of the UE and trunk rotation during the descent [50,217]. Ineffective UE responses to control forward landing and descent in older adults could be attributed to a number of factors including the response time, muscle activation, strength, range of motion or ability to absorb the force of impact and descent [20].

The task of successfully arresting a forward FOOSH is demanding in terms of speed, coordination and strength to prevent arm collapse [29]. The task can be divided into three general stages: movement of the arms during decent into a position in preparation for hand impact; impact of the hands; and energy absorption in the upper extremity (UE) to arrest the downward movement of the body [20]. While each stage is important, the current study focused on the final post-impact descent stage, which is a closed kinetic chain task, similar to the eccentric (lowering) phase of the common "push-up" exercise. Controlling the descent requires coordinated action of muscles spanning the shoulder and elbow (e.g., pectoralis major and triceps), combined with sufficient range of motion of these joints, and abdominal oblique muscle activation [173,174]. The pattern of muscle activation modulates the effective stiffness of the arm, ideally allowing significant angular rotation at the elbow and shoulder to absorb sufficient energy (to prevent head impact) without the production of high hand contact forces that may cause UE fracture [21-23,52].

Post-menopausal females tend to experience greater strength declines, decreased functional capacity, increased rates of sarcopenia, and increased risk for falls and fall-related injuries than their similarly aged male counterparts [26]. There has been little research examining differences between young and older women in neuromuscular and kinematic performance during a simulated forward FOOSH. We previously reported that older women were less able than younger women to absorb energy during a slow and voluntarily controlled forward descent on the outstretched arm [29]. While age-related reductions in elbow flexion and force production contributed, muscle activity and strength were not measured. Additionally, the slower speed of descent may not replicate the demands required when velocity increases. Other studies of muscular activity in younger men and women during a simulated fall arrest have focused on the muscles spanning the elbow (triceps brachii, biceps brachii) [23] and wrist (extensor carpi ulnaris and flexor carpi ulnaris) [218] and have not considered shoulder or trunk muscles. Muscle activation of the abdominals (external obliques) has been found to help stabilize the spine and prevent postural collapse when stability is challenged by an unstable surface [219], but female age differences in the activation of trunk muscles to control posture during a simulated post-impact descent has not been studied.

The purpose of this study was to determine the differences between young and older women in UE muscle strength, UE and trunk muscle activation and biomechanical factors during a task that simulated the final (post-impact) descent stage of a forward FOOSH at three body angles. Specifically, we hypothesized the older women would demonstrate decreased maximal elbow angle, energy absorption, elbow joint moment and increased force under the hand during a forward descent at three body angles. Additionally, we hypothesized that older women would have decreased UE concentric, isometric, and eccentric strength and decreased trunk activation when compared to the younger women.

3.2 Methods

We recruited young women aged 18 to 30 and older women aged 60 years and above. Women were excluded if they reported any of the following: a) a fracture to the wrist or forearm less than 2 years ago, b) any previous surgery to the UE, c) a recent (within the past 6 months) injury to the shoulder, wrist or hands, d) any current medical or neurological conditions involving weakness or pain in the UE, or e) any other recent significant medical or neurological concern (e.g., stroke, heart attack, chest pain). Exclusion criteria were determined utilizing a telephone screening questionnaire (Appendix A) and eligible women were scheduled for testing.

Each participant provided written informed consent (Appendix B), and the experimental protocol was approved by the University's Research Ethics board.

Participants completed all measures in one laboratory testing session. They first completed the Waterloo Handedness Questionnaire (WHQ) (Appendix C) and the International Physical Activity Questionnaire-Short Form (IPAQ) (Appendix D) [220]. Height and weight and limb lengths were measured. The testing session was then divided into two additional phases: strength assessment and forward descent test. The strength assessment was performed first for all participants.

3.2.1 Strength assessment

Strength was assessed using an Isokinetic Dynamometer (Humac Norm, CSMi, Stoughton, MA). Peak torque (Nm) was recorded during maximal effort isometric (ISO), concentric (CON) and eccentric (ECC) contractions of the non-dominant upper limb using a custom protocol developed to better simulate the plane of movement and muscle activation patterns required for a controlled body descent (Figure 3.1). This protocol was designed in order to estimate combined elbow and shoulder strength during a multi-jointed upper body movement task similar to the forward descent motion. For the ISO contractions, the shoulder was abducted 30° and horizontally adducted 45° and the elbow was flexed to 90° (Figure 3.1). During CON contractions, the participants moved the handle away from the body from a position of 60° of elbow flexion to full elbow extension. During ECC contractions, the participants initiated the contraction at a position of 120° elbow flexion and completed the contraction at 60° elbow flexion. The CON and ECC were standardized to 45 $°\cdot s^{-1}$ [0.78 $rad\cdot s^{-1}$]. Each test consisted of three maximum voluntary contractions separated by one minute rest periods. The order of contraction type was randomized. Each participant was given one or two practice repetitions before each contraction type. The standard encouragement was given for each test with the same tester for all participants. The position of the participant relative to the dynamometer axis of rotation meant that the torque values did not represent torque around a single joint (i.e. the elbow). Each test consisted of three maximum voluntary contractions separated by one minute rest periods. The custom protocol was pilot tested and reliability was confirmed for 10 older and 10 younger women. Test re-test reliability intra-class coefficients (ICC) over 3-5 days for both groups combined was: ISO, CON and ECC contractions r=0.932, r=0.907, and r=0.956 respectively.

Figure 3.1**.** Patient positioning during upper limb strength test. The participant was positioned in sitting with knees flexed at 90° over the edge of the seat, the posterior aspect of the pelvis

touching the back of the chair, thighs parallel with the edge of the seat, and the acromion process of the non-dominant side aligned with the edge of the back of the chair. The thigh/trunk angle was set at 90 degrees.

3.2.2 Forward descent test

In these trials, each participant performed a series of forward descents which resembled the downward portion of a push-up using a customized apparatus (see Appendix E). Each hand contacted a force plate (OR6-7, AMTI, Watertown, MA) that was mounted to a rigid, adjustable frame (Figure 2). The descent task was similar to the eccentric lowering phase of a push up and was designed to replicate the post impact phase of a simulated forward fall. The descent task was first described to the participants, and then demonstrated by the researcher and practiced three times against the wall. The task involved positioning the individual at a given body lean angle, and using a repetitive auditory stimulus to cue the onset and speed of their descent. This resulted in an approximate interval of one second between then start and end of the descent. Participants were told to start each trial with elbows in full extension without locking, shoulders flexed to 90° and hands shoulder width apart. They were encouraged to maintain a neutral spine and fully extend the knees without locking with feet maintained in the neutral positon and touching the foot plate. From the starting position, participants were told to lower themselves to 90° of elbow flexion while maintaining 30° of shoulder abduction (matching the strength testing protocol). If participants descended past 90° of elbow flexion the safety harness would stop the forward movement of their body to eliminate the chance of head impact. Participants began at the easiest level of difficulty (60° body lean) and then progressed to 45° body lean and the most difficult, 30° body lean. This sequence was chosen rather than a randomized order to ensure safety of not attempting a more difficult descent that would risk injury. The protocol consisted of one practice trial, followed by 3 repeated descents at each body lean angle, with one minute of rest between each repetition and a ten minute rest period between angles.

Before commencing the trials, measures were acquired of body height, weight and limb length. Arm length was measured from acromion to wall while the participant was standing with the shoulders flexed to 90° with hands flat on the wall. The foot length was measured from the lateral malleolus to the end of the longest toe. Shoulder height was measured from the acromion to the floor when the participant was standing in their usual posture. The height and angles of the force plates and the standing platform were then adjusted based on these measurements to ensure each participant's torso was parallel to the force plates with ankles in a neutral position on the standing platform, hands touching the force plate and arms fully extended for body lean angles of 30°, 45°, 60° from the horizontal. All participants wore a helmet with a full face guard and a safety harness secured to the ceiling by a tether that prevented contact of the head or torso with the force plates.

During each descent trial, an eight-camera, three-dimensional motion capture measurement system (VICON Nexus, VICON, Centennial, CO) was used to capture the positions of the surface markers at a sampling rate of 100 Hz. Forty-two reflective makers were used which enabled the calculation of 3D arm and body movement. These were located on the sides of the helmet, the front of the helmet, seventh cervical vertebra, tenth thoracic vertebra, fifth lumbar vertebra; and bilaterally at the acromion processes greater trochanters of the femurs, lateral condyles of the femurs, and lateral malleoli. Marker clusters were placed on the lateral distal shaft of the humerus and anterior proximal ulna. Surface markers were also placed over the sternum, and bilaterally over the lateral and medial epicondyles of the humerus and over the radial and ulnar styloid processes for the purposes of calibration. Kinematic data were low pass filtered at a cut-off of 15 Hz using a 4th order Butterworth filter. Shoulder joint centres were obtained through functional calibration methods [221] and arm kinematics were expressed using published standards [222]. Hand contact force was obtained from the force plates for both the right and left hands throughout the descent at each angle. Both the EMG and force plate data were captured at a sampling rate of 2000 Hz using an analog to digital board controlled by the motion capture system and were automatically synchronized to the kinematic data within the VICON software.

Electromyographic (EMG) data were recorded with a telemetered surface EMG system (Telemyo GT2400, Noraxon, Scottsdale, AZ). Muscle activity was recorded as mean amplitude value from the initiation to the cessation (peak elbow flexion) of the descent. Electrodes were placed on six muscle sites: anterior deltoid (AntDEL), pectoralis major (PM), triceps brachii (Long Head) (TRI), biceps brachii (BB), external oblique (EO) and internal oblique/transversus abdominus (IO/TrA). The electrodes were placed unilaterally on the non-dominant side. We chose to measure the non-dominant side only as this is the weaker side in a bilateral closed chain activity [223] and to limit testing burden for the participants. EMG signals were first recorded during three maximum voluntary contractions (MVC) using standard manual muscle testing positions [224-226]. MVC tests were all performed by the same researcher. One minute of rest was given between repetitions and the researcher gave standard verbal encouragement throughout each muscle contraction. Peak MVC EMG amplitude for each muscle was used to normalize EMG amplitudes arising in the subsequent descent trials.

The biomechanical variables evaluated in this study were: peak energy absorption (ENRG), normalized to height and weight; maximum vertical force (VF) in Newtons (N), normalized for body weight; maximum elbow flexion angle (FA) in degrees; and maximum elbow joint extensor moment (EM) in Nm/kg. ENRG is a measure of the total energy absorbed by upper limbs and was calculated using the vector sum of the total force applied to the hands (measured by the force plate) and the displacement of the shoulder (mean of right and left sides) [29]. VF was the maximum force observed during the descent. EM was calculated using standard inverse dynamics techniques and normalized by body weight. All data were calculated using custom software (Matlab, R2006b, Mathworks, Natick, MA).

A. B.

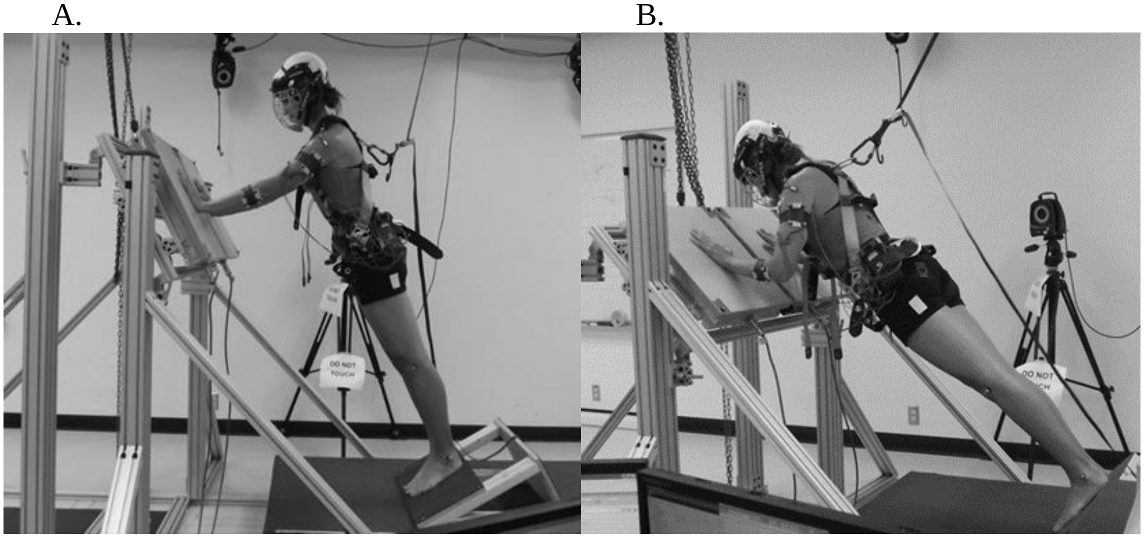


Figure 3.2 A. Starting (60° body lean) and B. end phase (45° body lean) of the descent on outstretched arms.

3.2.3 Data analysis

Statistical analysis was performed in SPSS 22.0 for Windows 8 (SPSS Inc., Chicago, USA). Independent t-tests were used to compare group differences in height, weight, total IPAQ score, FA and the time from initial motion to the end of elbow flexion (flexion time) at each descent angle of descent (60°, 45° and 30°) between age groups. This was done to determine if both groups were achieving the same velocity and depth of descent. Since FA is closely associated with ENRG, if there was a significant difference between the two groups, the FA was standardized for the calculation of energy absorption. Age differences in UE strength as measured by the dynamometer (i.e. ISO, CON, ECC) were determined using a repeated measures ANOVA with between subject factors (age (2) x within subject factors of contraction type (3)). Two separate mixed repeated measures multivariate analyses of variance (MANOVAs) (age (2) x body lean angle (3)) compared groups for muscle activation % MVC (AntDEL, PM, TRI, BB, EO, IO/TrA) and biomechanical variables (ENRG, VF, EM). When an overall multivariate age by body lean angle interaction was present, univariate interaction results were considered. If univariate findings were significant, further post-hoc analyses to compare age differences were conducted using independent t-tests at each angle of descent. Alpha was set at $p<0.05$. For repeated measures variables, Greenhouse-Geisser correction was used for violations of Mauchly's test of sphericity. Relationships of muscle strength to energy absorption was explored using Pearson r correlation coefficients.

3.3 Results

Participants included 20 healthy young women, age 18-30 years (mean 24.8±3.4), with mean body weight of 62.24±10.17 kg and mean height of 167.5±7.7 cm and 20 healthy older women, age 60 to 81 years (68.4±5.7), with mean body mass of 65.74±15.01 kg and mean height of 163.75±5.77 cm. All participants were right hand dominant according to the WHQ. All participants successfully completed the muscle strength testing and the descent trials at 60° and 45°. Two participants from the older group had difficulty performing the descent at 45° and did not attempt the 30°; therefore 18 older women were included in the muscle activation analysis.

An additional two older women did not reach 50° of elbow flexion during the descent at 30°; therefore, only 16 participants in the older group were included in the analysis for biomechanical measures during the descent tests. There were no significant differences in IPAQ scores, height or weight between the two age groups (Table 3.1). There was a significant difference in maximum FA achieved during the descent between groups. At each body lean angle younger women achieved a greater FA (Table 3.1) and FA was highly correlated to energy absorption (Pearson r correlation ranges from .64 to .83).

There was also a significant difference at each angle comparing groups; although both group means were close to the standardized time of 1 second. Flexion time was significantly higher in older women at each body lean angle (Table 3.1). Despite this difference, there were no significant correlations of descent speed to energy absorption (Pearson r correlations range from (-0.32 to 0.21).

3.3.1 Muscle strength

There was a significant interaction between age group and contraction type on muscle strength $F(2,76)=3.74$, $p=0.03$. Independent T-tests revealed that concentric strength was 15% greater in younger women compared to the older women $t(38)=2.25$, $p=0.03$. Torque values as measured by the Humac NORM can be found in Table 3.1. There were no significant correlations of any of the muscle strength measures to energy absorption. (Pearson r correlations ranged from -0.114-0.169

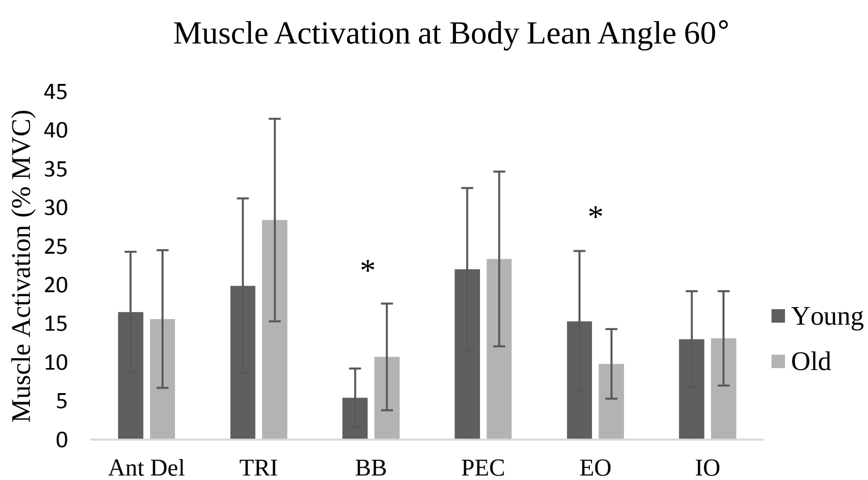
Table 3.1 Group characteristics including strength, elbow angle and flexion time during descent.

Independent T- Tests		p	Mean	Std. Deviation
IPAQ	Young		2987.70	2190.48
	Older	0.13	2139.31	994.28
Weight (kg)	Young		62.24	10.17
	Older	0.40	65.74	15.01
Height (cm)	Young		167.48	7.66
	Older	0.09	163.75	5.77
Strength				
ECC (Nm)	Young		89.95	30.25
	Older	0.80	91.70	17.37
CON(Nm)	Young		83.95	20.60
	Older	0.03*	71.21	14.90
ISO(Nm)	Young		75.00	32.73
	Older	0.65	71.05	19.56
Flexion Angle				
60°	Young		91.17°	6.87°
	Older	0.03*	84.95°	9.97°
45°	Young		81.63°	8.16°
	Older	<0.01*	73.15°	9.92°
30°	Young		75.67°	10.99°
	Older	<0.01*	61.99°	13.13°
Flexion Time				
60°	Young		1.03 sec	0.15 sec
	Older	0.01*	1.29 sec	0.26 sec
45°	Young		0.99 sec	0.12 sec
	Older	<0.01*	1.27 sec	0.26 sec
30°	Young		1.03 sec	0.17 sec
	Older	0.02*	1.23 sec	0.29 sec

Footnotes: IPAQ - International Physical Activity Questionnaire, ECC- Eccentric strength test, CON- Concentric strength test, ISO- Isometric Strength test. *Significant at $p<0.05$

3.3.2 Muscle activation

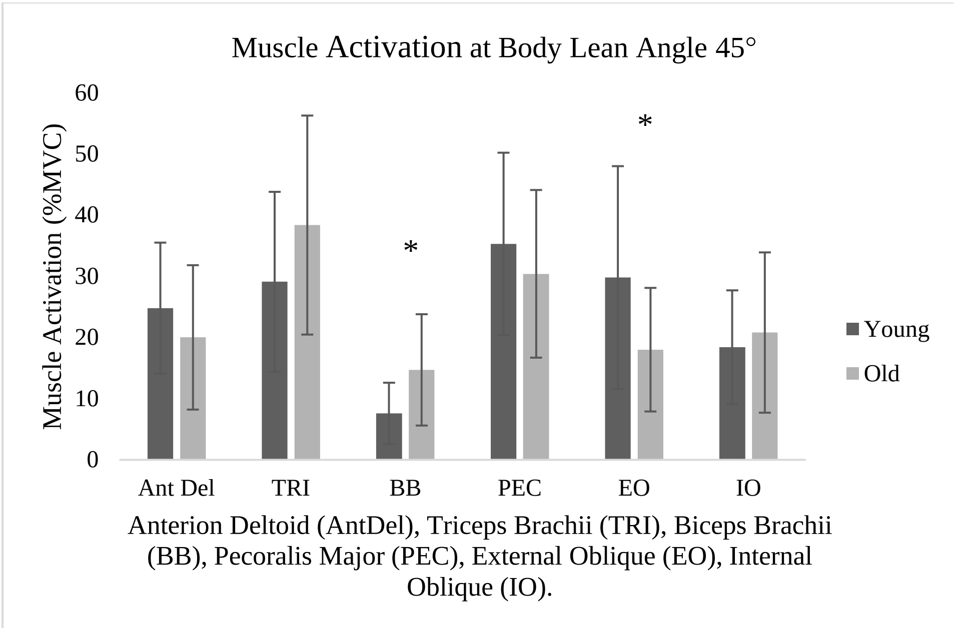
The multivariate analysis revealed a significant age group by angle interaction on muscle activity (*Pillai's T*(12,136)=3.39, *p*<0.01). Univariate tests revealed a significant group by angle interaction for BB *F*(2,72)=5.11, *p*=0.02 and EO *F*(2,72)=5.76, *p*<0.01 Further post-hoc analysis for the two univariate interactions (BB and EO), revealed that at all descent angles, 60°, 45° and 30°, older women had significantly greater mean BB activation *t*(38)=-2.91, *p*<0.01 *t*(38)=-2.87 , *p*<0.01, *t*(36)=-2.87, *p*=0.01, and less mean EO activation *t*(38)=2.26, *p*=0.04, *t*(38)=2.52, *p*=0.03, *t*(36)=3.35, *p* <0.01 compared to the younger group. At 30° of body lean, older women had 38% less EO activity than younger women (Figure 3.3).



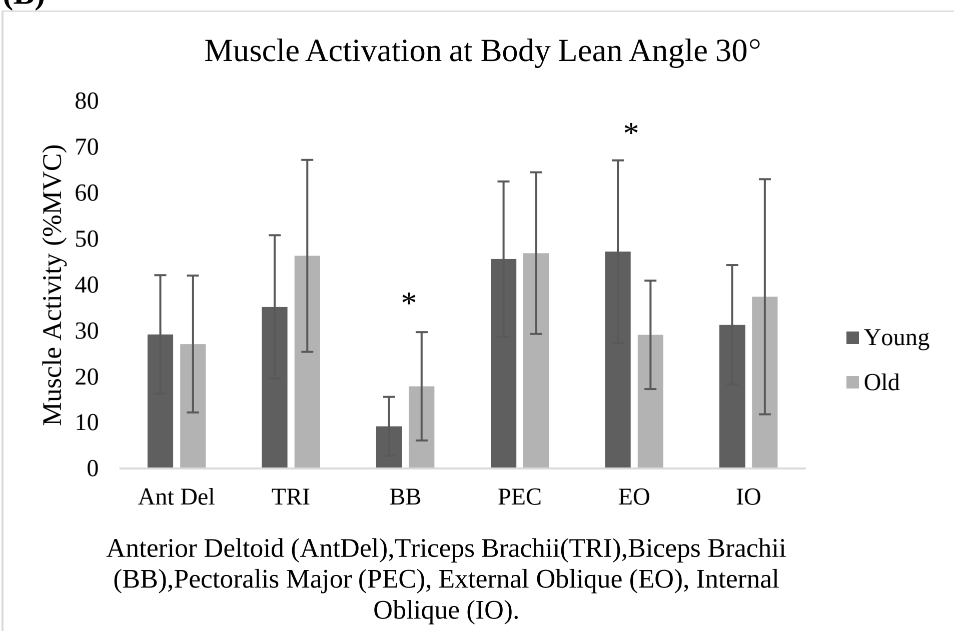
Anterior Deltoid (AntDel), Triceps Brachii (TRI), Biceps Brachii (BB), Pectoralis Major (PEC), External Oblique (EO), Internal Oblique (IO).

(A)

Figure 3.3. Comparisons of muscle activation between age groups at (A) 60° body lean angle, (B) 60° body lean angle, and (C) 30° body lean angle.
* Indicates significant difference p<0.05



(B)



(C)

Figure 3.3. Comparisons of muscle activation between age groups at (A) 60° body lean angle, (B) 60° body lean angle, and (C) 30° body lean angle.
* Indicates significant difference $p<0.05$

3.3.3 Biomechanical measures

After determining that FA differed between age groups, we chose to analyze ENRG at 50° of elbow flexion for both groups across all body descent angles as it was the highest amount of flexion that most women could achieve at each body lean angle (16 older women and 20 younger women reached 50°of elbow flexion during the descent at body lean angle 30°). This standardized the descent task by removing maximum FA as a confounding variable in the calculation.

The multivariate analysis revealed a significant age group by angle interaction on the biomechanical measures (*Pillai's* $T(6,134)=3.66$, $p<0.01$), with univariate tests revealing a significant age group by body lean angle interaction for ENRG $F_{GG}(2,68)=8.56$, $p<0.01$. Further post-hoc analysis using independent t-tests compared ENRG age differences at each angle found that at 30° body angle older women absorbed 19% less energy than the younger women $t(34)=2.59$, $p=0.01$. (Table 3.2)

Table. 3.2 Means and standard deviations of biomechanical variables

Outcome measures	YOUNG n=20		OLDER n=16	
Body lean angle 60°	**M**	**SD**	**M**	**SD**
Energy Absorption at 50° (Joules/% body weight x body height)	0.66	0.18	0.64	0.16
Max vertical force (% BW)	21.02	2.56	18.99	2.61
Elbow Joint Moment (Nm)	-2.05	0.39	-1.73	0.34
Body lean angle 45°				
Energy Absorption at 50° (Joules/% body weight x body height)	1.27	0.29	1.07	0.29
Max vertical force (% BW)	28.63	2.88	25.19	3.13
Elbow Joint Moment (Nm)	-2.19	0.73	-1.84	0.41
Body lean angle 30°				
*Energy Absorption at 50° (Joules/% body weight x body height)	1.97	0.45	1.60	0.37
Max vertical force (% BW)	35.14	2.95	32.98	3.22
Elbow Joint Moment (Nm)	-2.37	0.72	-1.94	0.45

Notes: %BW= % body weight in N

* Indicates significant difference $p<0.05$

3.4 Discussion

The purpose of this study was to examine the differences in UE strength, muscle activity and biomechanics during a FOOSH-like descent at varying body angles between younger and older women. We observed that older women had decreased UE concentric muscle strength, greater biceps and decreased abdominal muscle activation and decreased energy absorption compared to younger women. It is uncertain what the impact of these differences are on the risk of injury in a forward fall, but these factors may alter the ability to control the descent of the body post-impact when landing on outstretched arms in older women.

The demands of the controlled descent resulted in higher relative biceps muscle activity but lower relative levels of abdominal activation in older women compared to younger women. This occurred despite older women demonstrating preserved eccentric strength but lower concentric strength compared to younger women during a chest press motion. In the most challenging (30°) lean angle of our descent trials, older women had 19% lower mean values of energy absorption than younger women (even after controlling for elbow flexion angle). Our findings support the results of Sran et al. [29], who used a similar protocol and found that older women were able to absorb 45% less energy in the dominant arm than young women for body lean angles extending to 0° from horizontal. The collective results raise concerns about the ability of older women - even with relatively intact eccentric arm strength - to successfully absorb the impact energy of a fall in their upper limbs, and avoid head, torso or upper body impact.

Age-related loss of muscle mass, along with decreased isometric and concentric muscle strength have been well documented [92,227-229]. Older adults experience a 50% strength loss during isometric and concentric contractions compared to younger adults and eccentric deficits in strength have been observed at only 20% in men and absent in women [94-96,230]. These results support the decline in concentric strength in women, while eccentric strength was preserved. Pousson et al.[231] demonstrated that the older women's concentric (60°/s) elbow flexor strength was 54% of the younger women's. This study found that older women's concentric strength (45°/s) was 85% of the young women's. It is possible that higher speeds of contraction as used in Pousson et al.[231] may result in greater strength differences. Given that multiple muscles contribute to generating torque across several joints during a forward descent, this customized protocol may more accurately represent the strength requirements for arresting a fall; however further validation should be investigated. It was surprising that neither concentric nor eccentric muscle strength was correlated with energy absorption, suggesting that there is a complexity of factors influencing the ability to absorb the energy of forward descent. Additionally, the older women who participated in this study demonstrated similar levels of physical activity as compared to their younger counterparts. This may not be reflective of the typical older female population.

By examining muscle activity as well as energy absorption, our study provides new insights on the potential contributing mechanisms underlying age-related differences in the ability to control a body descent post-impact. At all lean angles, older women had greater activation of the BB during descent, but no difference in TRI activation, when compared to young women. The 49% BB difference between groups was consistent across lean angles. During the descent, the TRI activates eccentrically to facilitate elbow flexion motion while BB should exhibit relative inhibition in order for a controlled descent to occur. Although we did not measure co-activation directly, the observation of similar triceps activity, increased biceps activity and decreased elbow flexion angle achieved in older women suggests some degree of bracing or co-activation was occurring. In older adults, co-activation is most commonly described as a compensatory mechanism to increase joint stiffness [105,232]. Excessive stiffness of a joint can increase risk of bone injuries, while decreased stiffness can create instability and risk of soft tissue injury [172]. This strategy of co-activation is said to provide stabilization and to act as a braking mechanism [54] which some might argue provides some benefit in arresting a fall. On the other hand, co-activation can impair full activation of the agonist muscles via reciprocal inhibition causing diminished control of the body to the ground or support surface [233]. The age differences in BB muscle activation strategy may be one factor contributing to decreased energy absorption observed in older women, as well as to their inability to achieve the same degree of maximum elbow flexion. Optimally, co-activation of the antagonist might be decreased through strength training induced neurological adaptations [233].

It was hypothesized that older women would generate lower relative trunk muscle activation levels compared to younger women. This was supported; older women had significantly lower relative EO activation across all body angle descents. Traditionally, EO has been thought of as a trunk rotator; however, maintaining the spine in neutral requires bilateral isometric EO activation. Prone bridge exercises elicits higher levels of EO activity compared to supine trunk exercises [234]. Push-ups have been shown to involve high trunk muscle activity and therefore, they are used as a trunk training exercise because they provide a challenge to the abdominal musculature [173,174]. When trunk stability is challenged by an unstable surface, EO activation increases to stabilize the spine and prevent postural collapse [219], which then leads to biomechanical movement inefficiencies [235]. Postural collapse (difficulty maintaining a neutral spine and sagging into excessive lumbar lordosis) was observed in some of the older women in this study, but trunk angle was not measured directly.

The older women reached significantly less maximum elbow flexion during all body angle descents as younger women and performed the descent significantly slower. Although the descent time was not related to energy absorption, the maximum elbow flexion achieved was. However, even after controlling for elbow angle, energy absorption remained lower in older women. Voluntarily increasing the elbow flexion during forward descent impacts and while arresting an oncoming mass has been shown to decrease forward velocity, reduce impact forces and thus reduce potential fracture risk [20,22,23,40,41]. Although the women in this study were instructed to flex their elbows to 90° and perform the task at a standardized speed, the results achieved by the older women demonstrate a decreased ability to perform and control the standardized descent as instructed. It is difficult to speculate that the decreased concentric UE strength contributed to the differences in ENRG and FA since we observed no correlations between strength and ENRG. There are likely several contributing factors beyond the scope of this study.

Older women generated significantly lower energy absorption during the most difficult descent (30°) suggesting diminished ability to absorb the total energy in more demanding body positions that are closer to an actual fall. In a FOOSH, an optimal goal would be to decrease injury risk by reducing impact velocity with the hands prior to the head or the hip contacting the surface. Video analysis of falls has observed the use of the UE as the main mechanism to absorb energy during a descent in 84% of falls [63]. Because the older women could not descend as far with elbow flexion compared to the younger women, it raises concerns about their ability to successfully control a descent. Even when elbow angle was standardized across groups at 50°, older women continued to demonstrate diminished capacity to absorb energy. The current study furthers Sran et al.'s [29] findings by confirming decreased energy absorption occurs in older women even in their 60s and 70s. It also adds additional knowledge of muscle activation differences that may contribute to the diminished ability to control the descent, despite relative preservation of eccentric elbow strength.

One of the limitations of this study, although necessary to avoid excessive fatigue and test burden, was that measurement of strength and muscle activation was confined to the non-dominant UE. Choosing a specific target end range of elbow flexion for safety reasons could have detracted from the actual descent capabilities of the participants and not fully simulated the demands of arresting the body. Another limiting factor was the variation in the speed of descent in older verses younger women despite using a standardized protocol to control for this. Additionally, this lab protocol, where women control the descent with hands contacting the force plate does not simulate the other stages of fall descent such as impact.

In summary, we observed declines among older women in their ability to absorb energy during descent onto the outstretched hands, despite relatively intact eccentric UE muscle strength. Two potentially contributing factors, observed through EMG analysis, were increased BB co-activation (limiting effective eccentric elbow extensor activity for descent), and decreased EO activation (essential for stabilizing the spine). Findings from this study suggest exercise programs designed to address improving post-impact descent abilities should emphasize trunk stability training in older women. Multi-joint upper extremity strengthening may also be helpful to include in injury prevention programs, although further research needs to explore the relationship of UE strength to energy absorption and injury risk.

RELATIONSHIP OF STUDY 1 TO THESIS:

FOOSH 1 evaluated the simulation of a specific stage (Phase 3: Post Impact) of a FOOSH in young and older women. The purpose of study one was to compare muscle strength, muscle activity and biomechanical factors between young and older women during a controlled forward descent at 3 body angles. The post impact phase has important implications in arresting a fall in regards to preventing impact of the head or torso to the ground. By eliminating the impact phase of a FOOSH this study was safely able to investigate age differences during a controlled descent. This study established important age differences in biomechanical factors, and muscle activity during the post impact phase of a FOOSH; and age related strength differences in the UE. Evidence from this study suggests that older women executed a muscle activation pattern during the post impact descent phase that had decreases in muscle activity of the trunk (EO) and increases in BB activity when compared to younger women. It is hypothesized that this decrease in trunk activity could result in instability of the trunk which could affect the control of the descent. The increased activation of the BB during an eccentric triceps activity in the older women may be related to a strategy to create elbow joint stiffness that will be investigated further in study 3. Findings from this study highlight deficiency in older women's ability to control the descent phase and specific muscles to be targeted to include in injury prevention protocols to increase the ability of older women to control the descent during a FOOSH.

CHAPTER 4: STUDY 2

UPPER LIMB AND TRUNK MUSCLE ACTIVATION DURING AN UNEXPECTED DESCENT ON THE OUTSTRETCHED HANDS IN YOUNG AND OLDER WOMEN

Note: This manuscript is accepted to be published in the Journal of Electromyography and Kinesiology. The statement of copyright can be found in APPENDIX F. Some modifications have been made to accommodate the College of Graduate Studies and Research guidelines.

Journal of Electromyography and Kinesiology (2016), pp. 231-237

DOI information: 10.1016/j.jelekin.2016.08.001

4.1 Introduction

Falls are the leading cause of unintentional injury across all age groups in Canada with one third of adults over the age of 65 years sustain a fall annually [5]. Falls have a considerable negative physical, psychological and emotional impact on older adults as well as a costly burden on the health care system [7]. In Canada 85% of all injury related hospitalizations for older adults are the results of falls and the average cost per fall requiring hospitalization in Canada is $29 373.00 [44]. Age related declines in bone density, muscle mass, and neuromuscular reflexes may amplify the severity and occurrence of fall-related injury. In adults over 65 years of age, 60% of falls occur in a forward direction [47]. Landing on the hands, a protective mechanism to arrest the body and avoid injury to the head, has been observed in 74% of falls videotaped in long term care [48].

An unexpected landing on the hands during a fall requires the interaction of many factors including muscle activity, range of motion, speed, co-ordination, and hand placement. The muscle activity required to successfully arrest the body seems to be poorly understood. Dynamic restraint of joints in the body is achieved through preparatory and reflexive neuromuscular control [55]. In both expected and unexpected impacts, the central nervous system (CNS) must modulate muscle force before contact with a landing surface as well as throughout the time course of force application [54]. Fall arrest has been described as having two phases: pre impact and the descent phase (post impact) [20].

Pre impact muscle activity requires a motor control strategy to pre-set the muscles and joint positioning to absorb the impact [54]. The pre impact phase has been described as the phase where the neuromuscular system has the best advantage to configure the body in a manner to reduce the risk of fracture on impact [20]. Peak forces on the body have been observed almost immediately after impact [55,65,236]. The short duration of the impact phase, combined with the forces on the impact limb peaking within milliseconds after impact, would suggest that neuromuscular reflexes are not fast enough to substantially modulate the reaction forces during the impact phase [20]. This in turn leads to the deduction that the forces produced at impact are likely determined primarily by the neuromuscular activation patterns prior to impact [20]. The pre impact neuromuscular state during unexpected drop landings is better understood in the lower extremity where larger ground reaction forces (GRFs) are negatively correlated with pre-landing muscle activity patterns [54,160].

Post-impact muscle activity contributes to the later stages of deceleration and stabilization of body posture [54]. Post-impact modifications in muscle activity do not have the potential to reduce the peak impact force at the distal forearm [40] but likely play a role in diminishing the risk of impact to the head. Stretch reflexes are also likely to contribute to the muscle activity during the post impact phase [54].

Preparatory muscle activity has been found in the triceps brachii in young men when breaking a FOOSH [55]. Previous work confirmed the presence of preparatory muscle activation in the UE and deemed the anconeus, extensor carpi ulnaris and lateral head of the triceps as potentially important muscles in preparing for and effectively arresting a FOOSH [218]. During the 50ms interval prior to contact of the hand on the landing surface, DeGoede et al. [22] observed preparatory muscle activation in both the triceps brachii (51-81% maximum voluntary isometric contraction (MVC)) and biceps brachii (46-69 %MVC). The highest triceps muscle activity occurred in the “stiff arm” landings as opposed to the “natural” fall strategy in men. The highest preparatory biceps activity was during the minimal impact falls, where participants were instructed to “minimize the impact on their hands and catch the ground”.

Trunk muscle activation may also be important during a FOOSH because the trunk muscles create stability of the spine and thereby enable proximal to distal force generation that is thought to protect distal joints [237]. Trunk muscle activity also plays an important role in preparing for a fall impact through anticipatory postural adjustments (APAs) [238]. The ability to quickly modulate trunk muscle timing and recruitment in response to postural perturbations is considered paramount in maintaining balance and posture. Hodges et al. [181,193,198] found that feed forward recruitment of the transversus abdominus (TrA) preceded activation of all other muscles during movement of the upper and lower extremities. During self-initiated movements, APA activity is significantly delayed in the healthy older adults [200], as the postural muscles are recruited closer to or after the activation of the prime mover. Literature investigating trunk muscle activity during a simulated FOOSH is lacking.

Motor control of landing on the outstretched hands, particularly for older women, has important clinical relevance as failure to appropriately control impact absorption may lead to injuries to the musculoskeletal system [54]. There is evidence that older women sustain higher impact forces with less energy absorbing capacity compared to younger women during a forward descent [29,239]; however, it is not clear if this is due to neuromuscular activation differences or other factors. Understanding of age-related motor control differences of the pre impact and post impact phases could help explain why older women have a diminished ability to absorb the energy of a FOOSH.

The purpose of this study was to investigate the upper body and trunk muscle activity during a simulated unexpected FOOSH in healthy young and older women. We hypothesized that there will be differences in muscle activation between the young and older women. We hypothesized that young women will have greater amplitude of activation in the upper limbs and trunk prior to impact compared with the older women.

4.2. Methods

4.2.1. Participants

Young (age: 18-30 years) and older women (age: 60 years and above) were recruited. Exclusion criteria for this study were: a) fracture to the wrist or forearm less than 2 years ago, b) any previous surgery to the UE, c) recent (within the past 6 months) injury to the shoulder, wrist or hands, d) any current medical or neurological conditions involving weakness or pain in the UE, e) any other recent significant medical or neurological concern (e.g. stroke, heart attack, chest pain). Exclusion criteria were determined utilizing a telephone screening questionnaire (Appendix A) and eligible women were scheduled for testing. Each participant provided written informed consent (Appendix G), and the experimental protocol was approved by the University's Research Ethics board.

4.2.2 Data collection protocol

Participants completed the Waterloo Handedness Questionnaire (WHQ) (Appendix C) to determine limb dominance and height and weight were measured using a standardized protocol. The apparatus used in this study was designed to safely simulate the pre impact and immediate post-impact phases of a FOOSH. The limb length measurements were used to standardize the position of the force plates and foot platform (refer to a detailed description in 3.2.2). The participant was suspended in the apparatus with their body at a 60° angle from horizontal with their palms 1 cm above the force plates when their arms were outstretched and wrists extended (Figure 4.1). Participants wore a fitted safety harness as well as a full cage hockey helmet and were suspended from the ceiling by to a tether which was secured to an electromagnet-based quick release mechanism attached at hips. An additional ceiling mounted fall restraint cable was fixed to their harness to prevent any part of their body other than their hands from making contact with the apparatus. Participants were told to start each trial with the elbows in full extension, the shoulders flexed to 90° and the hands shoulder width apart. Participants were also encouraged to maintain a neutral spine with their knees in full extension and their feet touching or as close as possible in the frontal plane. From the starting position, participants were randomly released between 2-5 seconds after a verbal cue. Participants were instructed to "have a soft landing by using elbow flexion" and to try to descend to and not further than 90° of elbow flexion when they arrested the fall (Figure 4.2). Prior to testing with the apparatus, the FOOSH task was first described to the participants and demonstrated by the researcher and the participants were given three practice trials against the wall. Each participant performed 10 trials of an unexpected descent with their outstretched hands and the middle 5 trials were used for data analysis.

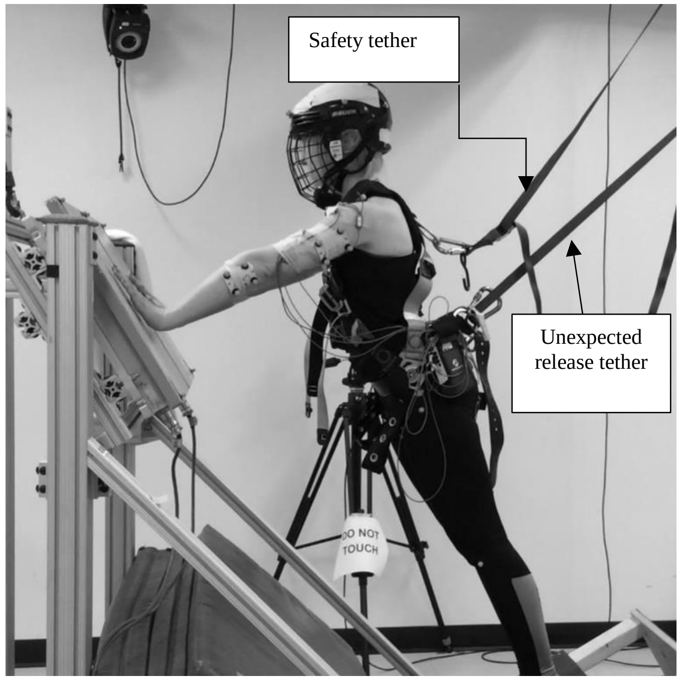


Figure 4.1. Participant hovering over force plates. (Baseline phase of FOOSH)

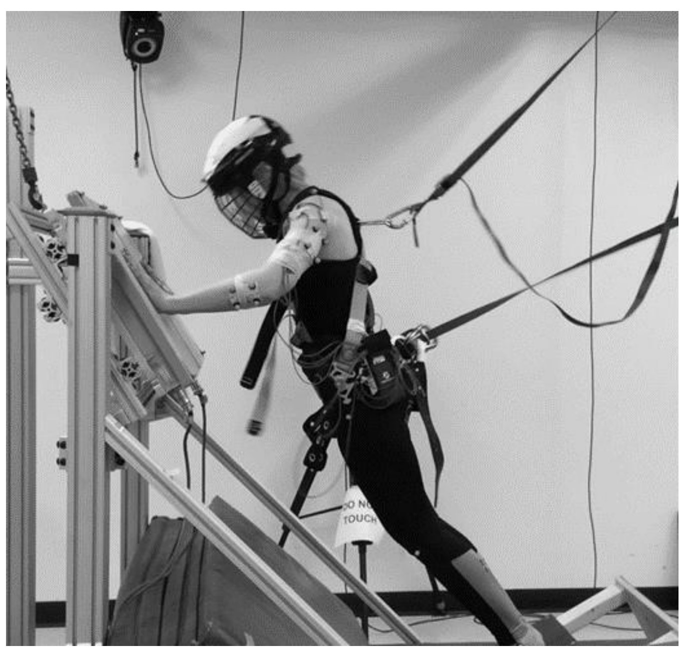


Figure 4.2 Participant post release (POST phase of FOOSH)

4.2.3 Instrumentation

Force data under each hand were collected using two force plates (OR6-7, AMTI, Watertown, MA) mounted to a rigid, custom built frame. Force data were sampled at 2000Hz. The height and angle of the platforms were adjusted to achieve the desired body lean angle based on height and limb length measurements. During each descent trial, kinematics of the upper body and arms were recorded using an eight-camera, three-dimensional motion capture measurement system (VICON Nexus, VICON, Centennial, CO) at a sampling rate of 200 Hz. A telemetered electromyography (EMG) (Telemyo 2400T G2, Noraxon) device with surface electrodes was used to measure muscle activity. Six pairs of Neuroplus Ag/Ag–Cl rectangular (2.54 cm^2) surface electrodes (Vermed, Bellows Falls, VT; A10043) were placed over the muscle bellies of each muscle in the direction of the line of action (20 mm inter-electrode distance).

Surface electrodes were placed on six muscle sites: anterior deltoid (AntDEL), pectoralis major (PM), triceps brachii (Long Head) (TRI), biceps brachii (BB), external oblique (EO), transversus abdominus/internal oblique (TrA/IO). Surface electrode placement sites were identified using standard protocols from previous research and skin was cleaned with alcohol to reduce skin impedance (Table 4.1). EMG electrodes were placed unilaterally on the non-dominant limb side as determine by WHQ. EMG signals were recorded during three isometric MVC attempts using standard manual muscle testing positions (Table1). EMG signals were recorded during maximum voluntary isometric contractions (MVCs) using standard manual muscle testing positions. The EMG signals had an active lead amplification of 500x with a 10 Hz high pass filter, CMRR >100dB and input impedance >100MOhm. Data were sampled at 2000 Hz and then filtered in software with a 20 Hz 10^{th} order high pass Butterworth filter and then full wave rectified and filtered with a dual pass 2^{nd} order low pass Butterworth filter with a cut-off frequency of 6 Hz to create a linear envelope.

EMG data were expressed relative to MVC taken as the peak value from three separate trials for each muscle group. Root mean squared EMG were calculated across the baseline phase (BL), defined as 500ms prior to release; the preparatory phase (PRE) defined as the individual's time between release and impact (group mean 257±37ms); and the post-impact phase (POST), defined as the 200ms after impact detected using the force plates. The means of 5 trials for each muscle was used for all statistical analyses.

Table 4.1. Muscle electrode placement, maximum voluntary contraction postures and methods.

Muscle	Electrode Placement	Participant posture	Participant action
Anterior Deltoid (An DEL)	One finger width distal and anterior to the acromion [240].	Seated with shoulder flexed with elbow flexed at 90° [241].	While tester stabilized scapula with one hand and applied a downward force on the distal humerus [241].
Pectoralis Major (PM)	On an angle midway between the anterior aspect of the humeral head and the nipple over the muscle belly [241].	Supine with shoulder flexed and abducted to 90° then externally rotated 15 ° with the elbow in slight flexion [242].	The tester resisted shoulder horizontal adduction and internal rotation [243].
Biceps Brachii (BB)	On the line between the medial acromion and the fossa cubit at 1/3 from the fossa cubit [240].	Sitting on plinth with the elbow flexed at a right angle and the dorsal side of the forearm in a horizontal downwards position [240].	Manual resistance applied against the forearm in the direction of extension [240].
Triceps Brachii (Long Head) (TRL)	Angled medial and inferior over the muscle belly. Half way distance from posterior acromion to olecranon and two finger widths medial of the line [240,241].	Seated on the plinth with the shoulder in 0° abduction, elbow flexed to 90° and forearm in 45˚ supination [240].	Extended elbow whole manual resistance was applied in the direction of flexion [240].

External oblique (EO)	Placed 15cm lateral to the umbilicus [244].	Participant lying supine with the knees flexed to 90° [245].	Tester resisted a curl up with trunk rotation towards the dominant side [246,247]
Internal oblique/ Transversus abdominus (TrA/IO)	Placed horizontally 2cm inferomedial to the anterior superior iliac spine within a triangle outlined by the inguinal ligament, lateral border of the rectus sheath and a line connecting the anterior superior iliac spine [244,248].	Lying supine with the knees flexed to approximately 90 degrees [224].	Participants were instructed to hollow their abdomen by drawing their navel up and in towards the spine. A Chattanooga pressure biofeedback device was placed under the lordosis of the lumbar spine, inflated to 40mmHg, with instructions for the participant to increase the pressure to 50mmHg [224].

4.2.4. Statistical analysis

Statistical analysis was performed in SPSS 22.0 for Windows 8 (SPSS Inc., Chicago, USA). Independent t-tests were used to compare group differences in height, weight and BMI. A group x time repeated measures MANOVA (2 age groups x 3 time phases: BL, PRE and POST) was performed to determine muscle activation changes over three phases comparing younger and older women for the following muscle groups: AntDEL, PEC, BB, TRI, EO, TrA/IO. If a significant multivariate interaction was found, univariate interaction effects were examined for each muscle, and Bonferroni adjusted multiple comparisons testing followed where appropriate. In addition, one-way repeated measures MANOVAs were conducted independently for each group to distinguish and plot time phase changes for each age group. If a significant multivariate effect of time was found for either group, univariate time effects and Bonferroni adjusted multiple comparisons were interpreted for each dependent variable. Alpha was set at $p < 0.05$.

Violations of Mauchly's test of sphericity were addressed using a Greenhouse-Geisser correction.

4.3 Results

Twenty young women (mean (SD) age: 22.9 (3.7) years; height: 168.9 (7.9) cm; body mass: 65.0 (8.7) kg; BMI: 22.9 (3.4) kg/m^2) and 20 older women (mean (SD) age: 68.1 (5.0) years; height: 161.5 (5.9) cm; body mass: 64.6 (11.2) kg; BMI: 24.7 (3.9) kg/m^2) underwent the unexpected descent protocol. There were no significant differences in weight or BMI between groups (p=0.899; p=0.110) and all participants were right hand dominant according to the WHQ. There was a significant difference in height between groups (p=0.02). This difference was accounted for in the methodology by having relative height adjustments for the equipment setup.

The multivariate analysis revealed there was a significant interaction effect of age group by time phase on muscle activity, Pillai's T(12,27),=4.05, p<0.001, a significant main effect of time phase on muscle activity, Pillai's T(12,27) =13.37, p <0.001, as well as a significant age group difference, Pillai's T(6,33) =3.59, p =0.008. Univariate tests revealed a significant group by time phase interaction only for IO/TrA F(2,76)=11.16, p<0.001) (Figure 4.3). The main effects of age group revealed younger women had significantly higher IO/TrA activity F(1, 38) = 8.31, p = 0.006. Further post-hoc analysis found a difference between ages only at IO/TrA PRE (p=0.001). Univariate tests revealed a main effect of time phase of the descent on muscle activity for all muscles measured pooled across groups; AntDEL, F(2,76)=12.08, p<0.001, PEC F(2,76)=53.72, p<0.001, TRI F(2,76)=46.24, p<0.001, BB F(2,76) =29.9, p<0.001, EO F(2,76)=43.09, p<0.001, IO/TrA F(2,76)=37.93, p<0.001; with Greenhouse-Geisser corrections used for AntDEL, TRI, EO and IO/TrA.

The one-way repeated measures MANOVA conducted separately for each age group, revealed that muscle activity of the AntDEL in the young had a significant increase from BL to POST ($p<0.001$). Muscle activity for the PEC and TRI progressively increased through the stages of the descent in both the young and old with highest activity in POST ($p<0.001$). In the young and old, BB activity significantly increased from BL to PRE ($p<0.001$) and the young's activity significantly decreased from PRE to POST ($p=0.003$). EO muscle activity increased significantly from BL to PRE for the young and old ($p<0.001$ for each). In the young and older women, IO/TrA activity significantly increased from BL to PRE ($p<0.001$ for each) but significantly decreased from PRE to POST ($p=0.006$) only in the younger women (Figures 4.4).

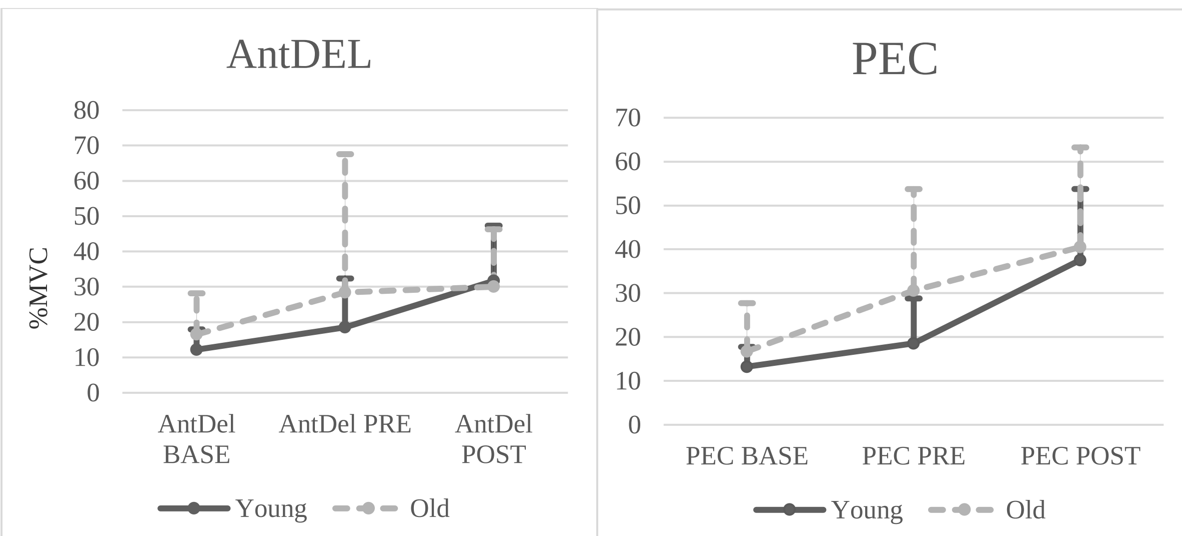


Figure 4.3 Mean muscle activity during three separate phases of a FOOSH (* indicates significant difference in muscle activity between age groups p=0.05)

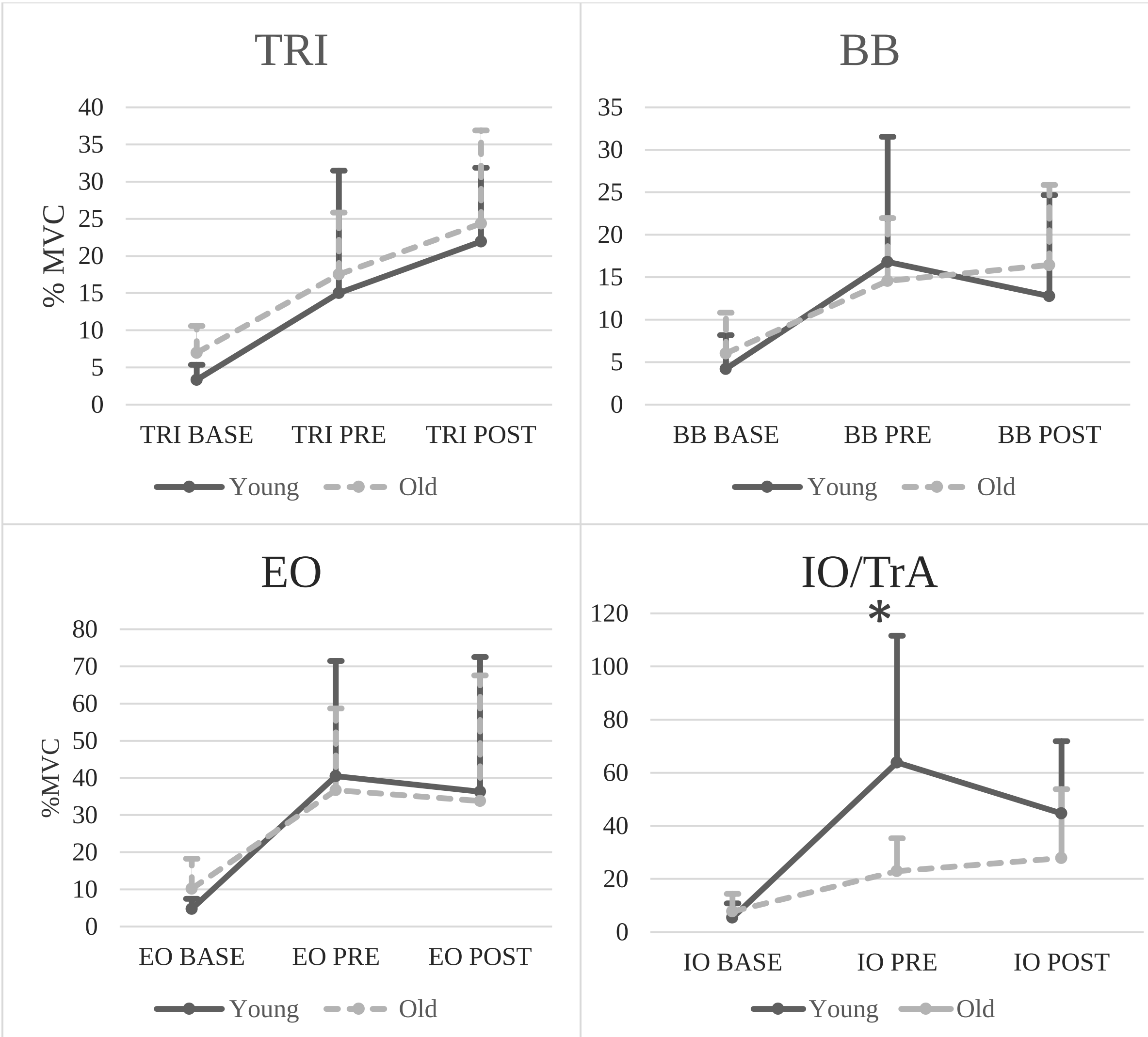
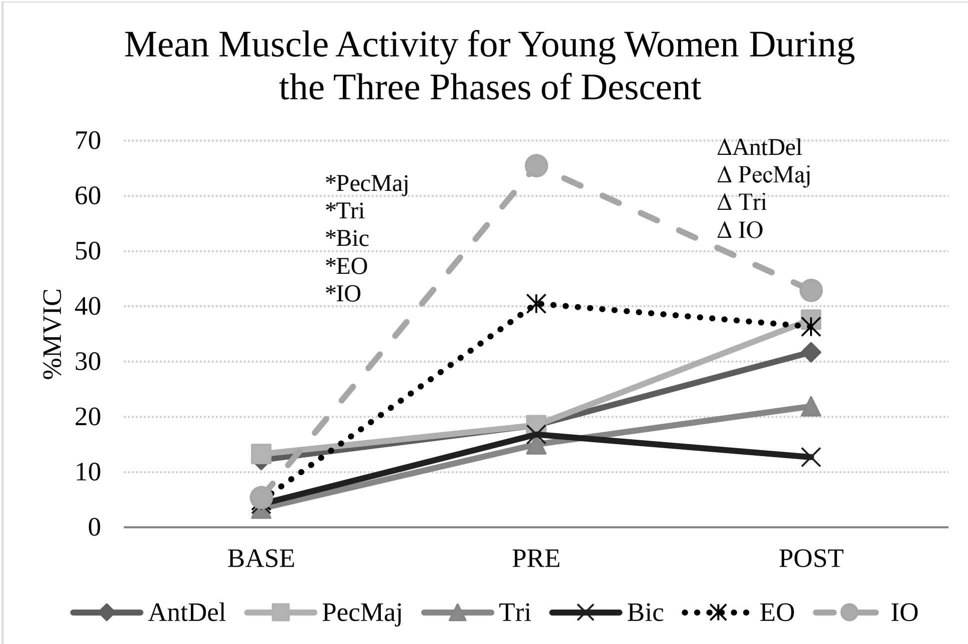


Figure 4.3 Mean muscle activity during three separate phases of a FOOSH (* indicates significant difference in muscle activity between age groups p=0.05)



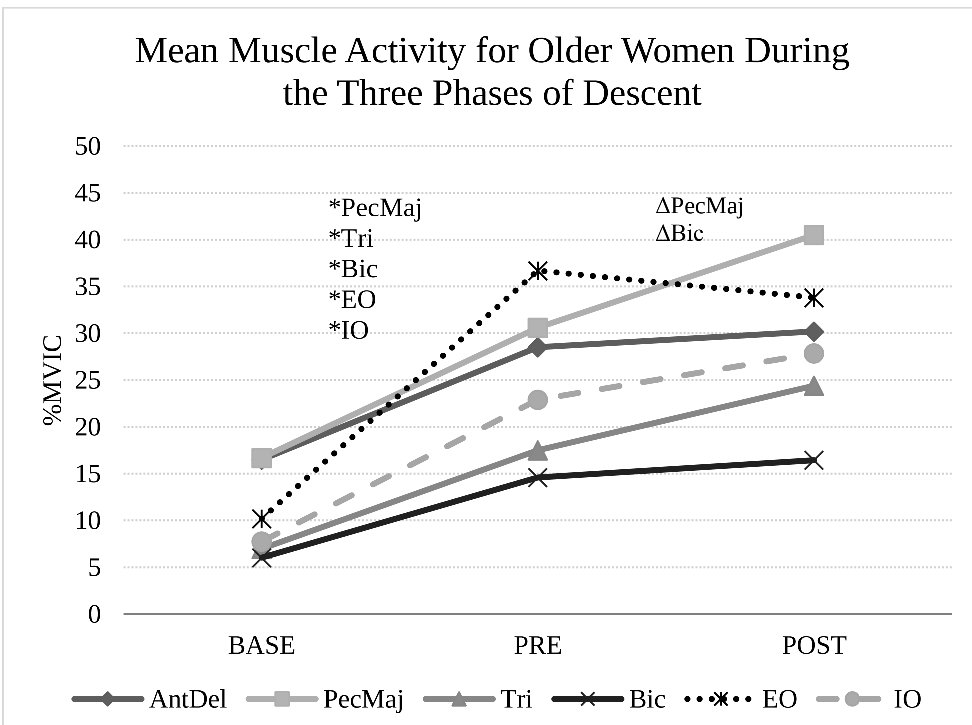


Figure 4.4. Mean muscle activity changes over time for young and old women. Significant changes between BASE and PRE phase indicated by * and Between PRE and POST phase indicated by Δ $p<0.05$.

4.4 Discussion

The objective of this study was to investigate muscle activity during the time period immediately before and after impact in an unexpected FOOSH in healthy young and older women. The main findings were that there were consistent, significant patterns of muscle activity in both younger and older women for the different phases of the unexpected descent; however, the primary distinguishing difference between the two age groups was that the younger women had significantly higher anticipatory (PRE) activation of the TrA/IO during an unexpected FOOSH.

It has been observed that both the TrA and IO activate in an anticipatory manner, meaning that the neuromuscular system recruits the TrA and IO prior to the prime mover during upper or lower limb movement as well as a response to perturbation [181 193]. Timely anticipatory activation of spinal stabilizers may help in minimizing potential disturbances to balance and maintaining postural stability during upper limb tasks. Anticipatory postural muscle activity (erector spinae and multifidus) is significantly delayed in healthy older adults compared to younger adults prior to upper limb loading [249].

Younger women demonstrated significantly increased IO/TrA activation during the PRE and decrease in the POST phase while the older women only increased activity of the IO/TrA during the PRE phase (Figure 4.4). Both young and older women had a similar pattern of EO activity, in that there is a significant change from BL activation to PRE and no significant change from PRE to POST. These data highlight the role of EO and IO/TrA as key trunk stability muscles for this task, possibly in an anticipatory role. The IO/TrA are known to be active prior to UE movements [181,198]. The recruitment of the IO/TrA and EO in the PRE phase could have been a mechanism to create a stiff proximal anchor against which the UE muscles can pull to generate torque and safely arrest the body without injury. Older women displayed a similar pattern (Figure 4.3) but the activity is at a much lower % MVC.

The IO and TrA are described as contributing to postural stability by increasing intra-abdominal pressure [250] and generating tension in the thoracolumbar fascia thus increasing spinal stiffness for intersegmental control. In this study, our unexpected FOOSH protocol did not involve a rapid limb movement similar to Hodges et al. [251] yet it was hypothesized that a perturbation in the form of being released from a supported position would also act as a stimulus to recruit an anticipatory response. To our knowledge, this is the first study to document feedforward muscle activity of the TrA/IO in anticipation of an unexpected FOOSH-like descent. Anticipatory activation and timing of activation of the trunk muscles may help to facilitate functional postural control. Older women may be more likely to show impaired trunk muscle responses, which could compromise the ability to stabilize the spine and increased the risk of injury [249]. Our finding of significantly reduced activation of TrA/IO in older women prior to upper limb landing (Figure 4.3) suggests that a decreased ability of the trunk musculature to maintain postural control, could lead to increased neuromuscular demand on other distal joint stabilizer muscles to absorb the landing force. Motor adaptive changes in anticipatory activity have been documented with training [252]. In experiments involving patients with low back pain, a single training session involving isolated voluntary contraction of the TrA muscle in supine position resulted in early APA onset in this muscle prior to arm flexion movements [252]. Voluntary TrA activation could be incorporated into trunk muscle training as part of a fall injury prevention program which is an exciting possibility for future research.

The difference in TrA/IO activation was the primary difference in muscle activation response between younger and older women; however, there were interesting differences observed in the timing of other UE muscle groups (Figure 2) that may suggest a shift of activation patterns with aging. All muscle measures with the exception of AntDEL, demonstrated significant increases in muscle activity from the BL phase to PRE for both age groups indicating anticipatory muscle activity. The lack of increased AntDEL activity observed may be explained by the positioning of the shoulder flexed to 90° in the BL phase. The muscle activity for the PEC and TRI progressively increased through the stages of the descent in both the young and older women with highest activity in POST. It is possible that older women may be increasing activation in the shoulder girdle to compensate for the reduction in trunk activation; but this needs to be explored in future research.

Proximal muscles such as the EO and IO are recruited in an anticipatory manner [181,193] as well as the antagonist (BB) to the prime mover (TRI). It appears that the young women adopt a muscle activity pattern that relies on the IO/TrA in an anticipatory fashion, and then once impact has occurred, trunk activity decreases with increasing activation at the shoulder girdle and elbow to control the descent. Younger women also demonstrated inhibition of BB as they significantly decreased activation in POST while the older women did not. It is possible that older women may be increasing activation in the shoulder girdle to compensate for the reduction in trunk activation; but this needs to be explored in future research.

The CNS controls movement and stability through both the feedback and feed-forward motor control mechanisms [73]. Looking at the overall muscle activation changes from PRE to POST we observed that the younger women had a significant decrease or increase in four out of six muscles examined possibly suggesting a modulation of muscle activity in a feedback manner. In contrast, muscle activation significantly increased in only two muscles in the older women (Figure 4). Older women have feed-forward activity in anticipation of a fall, but little modulation in muscle activity happens after impact. The limited muscle activity modulation might help to explain patterns of protective responses observed by Schonnop et al.[48] in real life video captured falls. They observed that older adults moved their arms into a protective position 74% of the time, yet the response was ineffective at preventing head impact during forward falls. The lack of modulation of post impact muscle activity could affect older women's ability to absorb energy during a forward fall since the POST muscle activity contributes to the later stages of breaking and stabilization of body posture [54].

A challenge of the study was the difficulty of simulating a natural forward fall in a controlled laboratory setting. Having the participants start with a flexed shoulder prior to an unexpected released was necessary for safety reasons, but unfortunately this does not fully represent a forward fall scenario. Although the current study did not test reactive arm positioning, it still gives insight into age related changes in muscle recruitment. Further examination should look at muscles linking the trunk and the upper extremity. Efficient force transfer could include larger trunk muscles such as the lattisimus dorsi. The lattisimus dorsi has direct communication between the EO, IO and TrA via the thoracolumbar fascia [253] and future research should include this muscle of interest.

This study describes upper body muscle activation strategies by young and older women to arrest the body during a FOOSH. Deficits in neuromuscular patterns prior to and after impact landing on the UE could potentially lead to unsuccessful falls on outstretched hands that could increase injury risk. The older women in this study did not activate the deep core muscles, TrA and IO, in the same anticipatory manor as younger women prior to a descent on outstretched hands nor did they demonstrate increased UE muscle activation after impact to control the descent as compared to younger women. Further study is needed, but this study provides support for targeting trunk and UE muscles in neuromuscular training programs designed to prevent fall related injuries for older women.

RELATIONSHIP OF STUDY 2 TO THESIS:

Study 2 and Study 3 utilize data from FOOSH 2, a comparison of a simulated unexpected descent on muscle activation and biomechanical factors in young verses older women. The phases of a FOOSH evaluated in this study included modified components of pre impact, impact, post impact and descent tested at a 60° body angle from horizontal. Study 2 focused on muscle activity during these phases whereas Study 3 will report the results of the biomechanical factors. Diminished activation of the trunk musculature to stabilize the spine may lead to biomechanical inefficiencies in movement and non-optimal force production of the upper extremities. Study 2 protocol allowed for the sequential examination of age differences in preparatory muscle activity as well as post impact muscle activity differences. The objective of this study was to investigate upper arm and trunk muscle activity during three phases; (Baseline (BL), Pre Impact (PRE), and Post Impact (POST) of an unexpected FOOSH in healthy young and older women. Motor control of landing on the outreached hands has important clinical relevance as failure to appropriately control impact absorption may lead to injuries of the musculoskeletal system. This study found that the primary differences observed were in the PRE phase where older women had significantly less activity of the IO/TrA. Younger women displayed preparatory activity of the BB and then significantly decreased activity POST while the older women did not. This suggests that older women may be utilizing a strategy of co-activation related to elbow stiffness and other biomechanical landing strategies which leads to the examination of these factors in study 3.

CHAPTER 5: STUDY 3

AGE DIFFERENCES AND MUSCLE STRENGTH RELATIONSHIPS TO LANDING BIOMECHANICS AND ARM STIFFNESS DURING AN UNEXPECTED RELEASE ON OUTSTRETCHED ARMS IN WOMEN.

5.1 Introduction

In Canada, falls are the leading cause of injury-related hospitalization for seniors and accounts for 32% of the $8.7 billion annual cost of injury [5]. In 2011 15% of the Canadian population was 65 years of age or older, where one third of this age group sustained a fall annually [254]. With the percentage of adults over the age of 65 years expected to double in the next 25 years [254], planning for an ageing population and preventing injuries from falls is a global priority [255].

Older adults living in the community self-report that approximately 60% of falls occur in a forward direction [47]. Reaching the hands to arrest the body's forward momentum is a protective response to avoid injury to the head, trunk or hip during a forward fall. An unfortunate cost to this protective response is that a fall on the outstretched hand (FOOSH) is the primary cause of fall-related upper extremity trauma [256]. Falls in a forward direction are frequently reported in community-dwelling older adults and hand impact during a fall is commonly observed in video surveillance data in long term care [20,47,48,216]. Of interest, head impact still occurred in 79% of the falls with hand impact [48]. An explanation of this finding could be that some older adults may not be able to effectively use the protective FOOSH response to prevent head impact. Post-menopausal women tend to experience greater strength declines, decreased functional capacity, increased rates of sarcopenia, and increased risk for falls and fall-related injuries than their similarly aged men [26]. Dynapenia or age related loss of muscle strength, could influence older women's protective responses [79] during a FOOSH with the resultant increased risk of UE fractures and traumatic brain injuries.

A FOOSH can be divided into three phases: 1) pre-impact, 2) impact and 3) post-impact deceleration. Pre impact phase is defined as the time period from loss of balance until impact [20] and involves muscle activity and UE movement strategies in preparation for landing. The second phase, defined as the impact of the extremity with the landing surface, occurs when forces act on the body peak within milliseconds and the energy of descent is absorbed through the extremities. The third phase is the post impact phase and involves muscle activity that contributes to the later stages of breaking the fall and stabilization of body posture [54]. Muscle forces are an important force attenuating component, which are able to actively adjust the amount of shock attenuation through eccentric contractions about the joint.

Elbow joint stiffness has been defined as the resistance offered by muscles and passive structures to deformation [257]. One of the primary modifiable neuromuscular factors that may contribute to elbow stiffness and controlling the post-impact descent is muscle strength. Excessive stiffness of a joint can increase risk of bone injuries, while decreased stiffness can contribute to increased instability and risk of soft tissue injury [172]. In older adults, co-contraction of muscles around the joint controlling descent (i.e. elbow) is most commonly thought to be a compensatory mechanism to increase joint stiffness [105,232].

During the second phase of a FOOSH impact forces reach 1-4 kN, which are enough to cause a wrist fracture in an older adult based on cadaveric studies [53]. Straight elbow or "stiff landings" produce greater peak impact forces, impulses, load rates and shorter impulse durations than self-selected or bent elbow landings [218]. Altering UE positioning prior to impact can reduce impact force by up to 58% where increased peak wrist impact forces are associated with greater shoulder flexion and less elbow flexion [56]. Impact force can be volitionally reduced by 32% by moderate flexion of the elbows (11°) prior to impact [22,64]. Chou et al.[40] found that if the elbow flexed upon impact this delayed the time to peak force and resulted in greater impulse. Flexion of the elbow at impact increases the dampening effect and could lead to increased energy absorption and thus decreasing the odds of impact of the head, hip or torso. Past studies have indicated that older women absorb 35%- 45% less energy during a controlled descent on the outstretched arms, but their energy absorption capabilities have yet to be investigated during an unexpected descent in women [29].

Previous work has demonstrated that elbow angle and wrist velocity at impact have a greater effect on peak impact force than do immediate post-impact adjustments in the UE such as post impact joint stiffness [56,258]. Controlling the post impact descent on the outstretched hands requires several coordinated muscle actions at the shoulder and elbow in order to control the descent to lower the body to the ground or landing surface. Although there is some evidence where older women are less able to flex their elbows to descend as far as younger women can in a controlled reverse push-up motion [29,239], it is unclear what differences exist in this phase of a FOOSH when the descent is unexpected.

Determining the mechanisms that may affect injury risk would be helpful to guide the development of interventions to reduce risk and severity of injuries. Studies involving quantifying landing strategies in women is limited. Past in vivo laboratory studies of forward falls have investigated fall strategies from varying heights in male participants or young adults where descents were controlled or participants were aware of the fall [40,41,52,56].

The objectives of this study were to compare age differences in biomechanical and physiological variables that contribute to fall arrest strategies in an unexpected descent on outstretched arms and determine the association of UE muscle strength to these variables. It was hypothesized that: 1) older women would exhibit decreased muscle strength in the UE compared to younger women, 2) older women would demonstrate a fall arrest and descent strategy with decreased elbow flexion angles at impact and descent phases, decreased energy absorption and greater elbow stiffness and 3) UE strength would be associated with energy absorption for both older and younger women. Other biomechanical variables likely to contribute to injury risk such as impulse duration and velocity at impact were also explored, but no hypothesis was set as potential age differences are not clear from previous literature.

5.2 Methods

Participants were recruited by community poster and newspaper advertisements. Potential participants were screened for eligibility with a telephone interview (Appendix A). Exclusion criteria for this study were: a) fracture to the wrist or forearm less than 2 years ago, b) any previous surgery to the UE, c) recent (within the past 6 months) injury to the shoulder, wrist or hands, d) any current medical or neurological conditions involving weakness or pain in the UE, e) any other recent significant medical or neurological concern (e.g. stroke, heart attack, chest pain). Written informed consent was obtained, and the experimental protocol was approved by the institution's Biomedical Research Ethics board.

5.2.1 Data collection protocol

Participants first completed the Waterloo Handedness Questionnaire (WHQ) (Appendix C) and the International Physical Activity Questionnaire-Short Form (IPAQ) [220] (Appendix D). Height and weight were measured using a standardized protocol with a portable stadiometer and a weigh scale. Additionally, limb length measurements were used to standardize the position of the force plates and foot platform (refer to a detailed description in 3.2.2).

5.2.2 Strength assessment

Strength was assessed using an isokinetic dynamometer (Humac Norm, CSMi, Stoughton, MA) using the same protocol described in 3.2.1 and depicted in Figure 3.1.

5.2.3 Unexpected descent

Participants performed an unexpected FOOSH as described in Section 4.2.2. and depicted in Figures 4.1 & 4.2. A telemetered electromyography (EMG) (Telemyo 2400T, Noraxon) device with surface electrodes was used to measure muscle activity. Two Neuroplus Ag/Ag–Cl rectangular (2.54 cm^2) surface electrodes (Vermed, Bellows Falls, VT; A10043) were placed over the muscle bellies of each muscle in the direction of the line of action (20 mm inter-electrode distance). Surface electrodes were placed on two muscle sites biceps brachii (BB) and triceps brachii (TRI). EMG signals were recorded during three maximum voluntary isometric contractions (MVC) using standard manual muscle testing positions. BB was tested using manual resisted elbow flexion with the participant seated on the plinth with the shoulder in 0° abduction, elbow flexed to 90° and forearm in full supination [259-261]. Positioning for TRI testing was the same as BB, except with forearm in 45° supination, resisting elbow extension [240]. EMG data from each trial were expressed relative to MVC taken as the peak value from three separate trial for each muscle group.

During each descent trial, 3D kinematics of the upper body and arms were recorded using an eight-camera, three-dimensional motion capture measurement system (VICON Nexus, VICON, Centennial, CO) at a sampling rate of 100 Hz. Forty-two reflective makers were used which enabled the calculation of 3D arm and body movement. Elbow and shoulder joint centres were obtained through functional calibration methods [221] and UE kinematics were calculated using published standards [222]. The EMG had an active lead amplification of 500x with a 10 Hz high pass filter, CMRR >100dB and input impedance >100MOhm. Data were sampled at 2000 Hz, first filtered with a 20 Hz 10^{th} order high pass Butterworth filter and then full wave rectified and filtered with a dual pass 2^{nd} order low pass Butterworth filter (cut-off frequency of 6 Hz) to create a linear envelope. Root mean squared EMG were calculated across 100ms prior to impact and 200ms post impact on the force plates. Both the EMG and force plate data were captured at a sampling rate of 2000 Hz and were synchronized to the kinematic data.

The elbow flexion angle of the left arm was calculated and referenced to the angle at the time of release. The value of the elbow angle at impact (ImA), the elbow angular velocity at impact (ImV) and the elbow angle at 200 ms post impact (EnA) were extracted. Peak energy absorption (ENRG) and total impulse (ImP) were calculated from the data collected 200ms after impact. ENRG is a measure of the total energy absorbed by the body and is calculated as the integral of the dot product of the reaction force and the displacement of the trunk [29]. ImP was the integral of the total force magnitude measured under the left hand over the first 200 ms after contact and normalized to body mass. Elbow stiffness (ES) was calculated as the slope of the linear regression relating elbow flexion angle and elbow flexion moment (calculated using standard inverse dynamic techniques) during the first 200 ms after impact. A more negative value represents a higher degree of stiffness. Co-contraction ratios of the BB muscle activity with respect to the TRI activity prior to hand impact (Co-acPRE) and post impact (Co-acPOST) were calculated [262,263]. To obtain the Co-acPRE, the root mean square (RMS) amplitudes (normalized to MVC) of the BB and TRI were obtained over the 100 ms prior to hand impact. For Co-acPOST, the BB and TRI RMS amplitudes were calculated over the 200 ms post hand impact. All data were calculated using custom software (Matlab, R2006b, Mathworks, Natick, MA).

5.2.4 Data analysis

Statistical analysis was performed in SPSS 22.0 for Windows 8 (SPSS Inc., Chicago, USA). Independent t-tests were used to compare group differences in height and weight and IPAQ scores. Age differences in UE strength as measured by the dynamometer (ISO, CON, ECC) were determined using independent t-tests with a Bonferroni correction $p<0.016$. A multivariate analysis of variance (MANOVA) test determined age differences in biomechanical variables and UE muscle activity (ImA, ImV, ENRG, EnA, ImP, ES, CO-acPRE, CO-acPOST). Multivariate tests (Pillai trace) were considered significant at $p< 0.05$. Univariate ANOVAs were used for post-hoc testing when significant between group differences were found in the MANOVA. Pearson correlations were run to determine the relationships between UE strength (ISO, CON, ECC) and biomechanical variables (ImA, ImV, ENRG, EnA, ImP, ES) in each age group. Alpha was set at $p< 0.05$. Violations of Mauchly's test of sphericity were corrected using a G**ree**nhouse-Geisser correction.

5.3 Results

Nineteen healthy young women, aged 18-30 years (mean 23.0±3.8 [SD]), with mean body mass of 64.8±8.9 kg, mean height of 169.1 ±8.2 cm and IPAQ score 4889.9± 2952.8 were included in this study. Eighteen older women included in this study were aged 60 to 78 years (68.3±5.4), with a mean body mass of 64.4±8.8kg, mean height of 160.9± 5.8 cm and IPAQ score 3604.3±2197.1. There were no significant differences in IPAQ scores ($p=0.116$) and no significant differences in weight between groups ($p=0.866$). There was a significant difference in height between groups (*p*=0.02). Adjusting the position of the equipment relative to each participant's height controlled for this difference. All participants were right handed according to the WHQ.

5.3.1 Muscle strength

The older women had significantly less elbow extension concentric strength (CON) compared to the younger women *t(*33)=3.39, p=0.002; whereas isometric t(33)=2.15, p=0.039 and eccentric t(33)=0.635,p=0.53 were not significantly different (Figure 5.1)

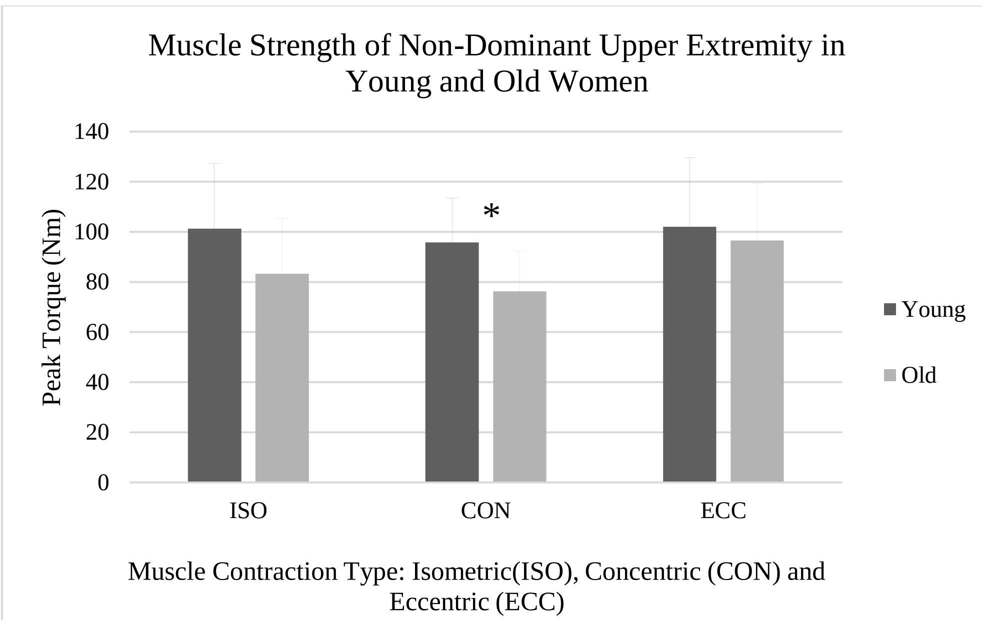


Figure 5.1. Muscle Strength comparisons between age groups across contraction types. Significant muscle strength differences between age groups indicated by *

5.3.2 Unexpected descent

The multivariate analysis revealed a significant multivariate effect of age group on biomechanical measures using *Pillai's T*(5,29)=3.04, *p*=0.005. Further post-hoc analysis found that young women had significantly greater ImV and EnA compared to the older women

$F(1,33)=8.54$, $p=0.02$, $F(1,33)=5.83$, $p<0.001$. Younger women were able to absorb 36% more ENRG with their UE when compared to the older women $F(1,33)=13.55$, $p=0.001$ (Table 5.1).

Table. 5.1 Means (M) and standard deviations (SD) of biomechanical variables comparing impact and post impact forward descent in older and younger women.

Biomechanical Variable	*p* value	M Young	SD (±) Young	M Older	SD (±) Older
Impact Angle (ImA) (°)	*0.006	14.6	14.1	3.6	5.8
Impact Velocity (ImV)(°/sec)	*0.02	118.37	107.32	42.87	67.08
Elbow Joint Stiffness (ES)(N/m)	0.47	-0.27	0.071	-0.29	0.086
End Elbow Angle (EnA) (°)	*0.00007	58.1	17.4	33.1	14.46
Impulse (IMP) (% BW in [N s])	0.36	0.275	0.058	0.25	0.087
Energy Absorption (ENRG) (Joules/body weight x body height)	*0.001	0.011	0.003	0.006	0.002
Coactivation PRE (Co-acPRE) (%MVC BB/ %MVC TRI)	0.51	1.29	0.66	1.11	0.94
Coactivation POST (Co-acPOST) (%MVC BB/ %MVC TRI)	0.16	0.61	0.4	0.94	0.89

Significant differences between age groups indicated by *.

5.3.3 Strength and biomechanical correlations

In the younger women CON was positively and moderately correlated [264] with ENRG $r=0.492$, $p=0.032$ and EnA $r=0.538$, $p=0.017$. In the older women, all muscle contraction types were moderately positively correlated with energy absorption (ECC r=0.577, $p=0.019$, CON $r=0.668$, $p=0.005$, ISO $r=0.576$ $p=0.02$. In the older women eccentric strength was also positively moderately associated with ImV $r=0.508$, $p=0.045$ and ES $r=0.514$, $p=0.042$. (Table 5.2)

Table 5.2. Correlations between strength measures and biomechanical variables in both young and older women

Young Women	Impact Angle (ImA)	Impact Velocity (ImV)	Elbow Stiffness (ES)	Energy Absorption (ENRG)	Impulse (ImP)	End Angle (EnA)
Eccentric (ECC)	0.167	-0.275	-0.208	0.414	0.432	0.14
Concentric (CON)	0.423	0.302	-0.124	0.492*	0.035	0.538*
Isometric (ISO)	0.123	-0.045	0.156	0.262	0.288	0.156
Older Women						
Eccentric (ECC)	0.191	0.508*	0.514*	0.577*	0.256	0.492
Concentric (CON)	0	0.206	0.04	0.668**	0.325	0.436
Isometric (ISO)	0.081	0.205	0.109	0.576*	0.03	0.493

** correlation is significant at p<0.01

* correlation is significant at p<0.05

5.4 Discussion

The primary objectives of this study were to compare biomechanical and physiological variables related to forward fall arrest strategies during an unexpected release and UE muscle strength between young and older women. A secondary objective was to explore the relationship between UE strength and biomechanical parameters associated with an increased risk of fall-related injury. The main findings of this study supported the hypotheses that age-related differences exist in both elbow extensor strength and forward fall arrest and descent strategies in women. These differences may contribute to increased risk of injury during a FOOSH. Specifically, older women had decreased elbow flexion and angular velocity at impact compared to younger women representing a distinct age difference in landing strategy. Of importance to injury risk, older women are not able to absorb as much energy on impact as younger women, which may increase the risk of injuries such as wrist fractures or head impact. The reasons for the difference in energy absorption are not completely clear, as the older women demonstrated relatively preserved UE ECC strength.

During a FOOSH, an optimal goal would be to absorb enough energy with the UE to reduce the risk of impact to the head or torso. It has been observed that older women absorb 45% less energy with their UEs at body lean angles up to 90° from vertical compared to younger women [29]. In a previous study by our team, when elbow angle was standardized to 50° across groups and participants were at a 45° body lean angle, older women continued to demonstrate 18 % less energy absorbing capacity than younger women (Lattimer et al. 2016; refer to Chapter 3). The pre-impact and impact phases were not measured in this controlled FOOSH protocol and, therefore do not include the increased rate of stretch anticipated in the eccentrically contracting muscles. This study, the first to our knowledge to investigate an unexpected descent in older and younger women, adds to these studies by finding an even greater loss in energy absorbing capacity in older women when the increased rate of stretch increases the demands for muscle force generation. This was observed despite a relative preservation of eccentric strength in the older participants.

The amount of energy absorption by a joint can be affected by ImV or resultant joint moment. In the lower extremities (LE) greater energy absorption has been observed during landings with greater angular displacement at the hip, knee and ankle. In the LE, the angular velocities of the hip and knee at the moment of foot impact significantly affect the peak ground reaction forces [265]. Yu et al.[265] postulated that active flexion of the proximal joints reduces the impact force at the distal point of impact. In this study, the younger women had significantly greater elbow angular velocity, representing a quicker active elbow flexion motion prior to impact which may have contributed to greater energy absorption.

Body positioning and velocity at impact may influence whether a forward fall results in injury [20]. Risk of fractures in the UE is thought to be influenced by impact velocity and elbow angle. [40,266] Specifically, DeGoede et al.[266] observed a 0.9% decrease in impact force for each degree increase in elbow flexion. The present study observed a mean difference of 11.2° of elbow flexion on impact between the older and younger women, which may translate into a 10% increase in peak force at the wrist according to DeGoede et al.[57]

Greater ImA places the triceps muscle in a more lengthened position and it is known that changes in muscle length affect muscle activity. In young adults, the highest muscle strength and activity was found at 84° of elbow flexion compared to lesser angles as measured by dynamometry [267]. In the lower extremity, older adults have a decreased capacity to develop eccentric extensor torque at a stretched muscle length compared to shortened [268]. Perhaps, the older women's ability to generate force would have been compromised at greater elbow angles and they were operating at their optimal muscle length, which was less than that of the younger women.

Age-related loss of muscle mass, along with decreased isometric and concentric muscle strength, has been well documented [227]. Our results support that the reduction in elbow extensor concentric strength in older women is more pronounced than reduction in eccentric strength. The preservation of eccentric strength in older adults is supported in the literature [92,229]; however, there has been limited study of UE strength. In the older women there was a moderate correlation [264] with all of the strength measures and ENRG. This suggests that strength deficits in the UE might impact the energy absorbing capacity of the UE in older women to a greater extent than younger women. In older women, eccentric strength was positively correlated with impact velocity of the elbow and elbow stiffness. This suggests that there may be an association between eccentric strength and landing strategies in older women that is not apparent in younger women. The relationship of all UE elbow extensor strength measurements to energy absorption during a FOOSH in older women may point to the importance of maintaining and increasing UE strength as women age. Our novel strength testing protocol was multi-joint and is a more complex movement pattern than single joint isokinetic testing. Perhaps practicing complex UE movement patterns against resistance could be beneficial to injury prevention programs. Perhaps focusing on the eccentric phase of exercises such as "push-ups" would familiarize the movement patterns and increase strength in fall specific patterns.

The triceps brachii muscle group is the primary contributor to elbow stiffness when the joint is forcibly flexed [167]. Little is known about how arm muscle activity contributes to elbow joint stiffness in a fall arrest scenario in women. Co-contraction is associated with recruitment of both agonist and antagonist and increases rotational stiffness. The current study found no age differences in the co-contraction ratio either pre- or post-impact. The ECC test, of the UE strength tests employed, most closely resembles the lengthened position of the triceps while the elbow is being forcibly flexed by a fall impact. Interestingly, we found no age differences in ECC strength or in the co-contraction ratio of the TRI to the BB, suggesting that older and younger women do have similar muscle co-activation capabilities to arrest the body during a FOOSH. Given the high variability of the data, it is possible that a larger sample may have found significant differences. The trend from our mean data revealed decreased co-activation at impact and increased co-activation post-impact in older women compared to younger, although this was not significantly different. Lee et al.[167] found that elbow extensor pre-activity translated to greater resistance to stretch in men than in women. Lee et al. [269] concluded that, at the same volitional co-contraction levels, the reason for the gender differences in stiffness and dampening was greater elbow extensor strength in men.

Elbow stiffness varies with movement speed, joint positioning and muscle contraction level [270]. Kuxhaus et al.[270] found that elbow stiffness can be influenced by rotational joint speed (ImV): increased speed increases joint stiffness. Since this study found no significant age differences in ES, it is possible that the two groups had different tactics for creating stiffness. For example, younger women had increased ImV and ImA, while older women created the stiffness with similar muscle activation levels. Both age groups had similar muscle activation, yet different UE postures on impact. One possible explanation for this may be that the younger women were activating their BB and TRI to actively flex the elbow to absorb impact on landing, while the older women were activating their BB and TRI to create joint stiffness in a more extended elbow position.

To safely include older women in our study, we chose to include the instructions of "have a soft landing by using elbow flexion". This may have changed the natural fall arrest strategy of our participants but was considered necessary for safety reasons and to avoid injury. Having the participants in a flexed shoulder position prior to an unexpected released does not fully represent a forward fall scenario. Perhaps, changes to the protocol to more closely replicate a natural fall scenario would highlight other UE reactive differences between age groups.

In conclusion, the objectives of this study were to compare biomechanical variables of fall arrest strategies and muscle strength in the UE between young and older women. Despite having no differences in elbow stiffness, the older women absorbed far less energy with their UEs than the younger women. Fall arrest strategies were different between age groups; younger women demonstrated increased elbow velocity and flexion angle at impact and a much greater end elbow flexion angle during descent. This strategy appears to be optimal to increase the amount of energy absorbed by the UE during a simulated FOOSH. In the older women, UE strength was associated with energy absorption capacity, and they had significantly less concentric strength than the younger women. The modulation of energy absorption capabilities by altering elbow velocity, increasing elbow angles at impact, as well as increasing UE strength are all potential factors that could be modified through training in order to decrease potential injury risk.

RELATIONSHIP OF STUDY 3 TO THESIS:

Study 3 involved the same FOOSH 2 cohort as Study 2 with the primary aim to compare biomechanical variables of fall arrest strategies (unexpected) and muscle strength in the UE between young and older women and to evaluate the association between UE strength and the biomechanical variables. This study highlighted distinct age related differences in fall arrest strategy during a simulated fall on outstretched hands. This study provided evidence of age related fall arrest differences where younger women had greater elbow angles and elbow velocity at impact. This strategy difference suggests that younger women are controlling their arms and landing with a slightly flexed elbow, mimicking the landing strategy proposed to decrease the impact force as identified in earlier studies [22]. Older women are doing the opposite; in order to arrest the fall, they did not actively flex their elbows at impact to the same extent, despite instructions to do so, and this resulted in a decreased ability to absorb the energy of the body. This strategy could translate to a decreased likelihood of arresting and descending the body in a controlled manner and, thus, could increase the likelihood of injuries. This study also added to the findings from Study 1 during the controlled post impact phase, confirming that older women were able to absorb significantly less energy upon impact with the UE in an unexpected descent. The results of this study also corroborate the decreased UE strength found in the first cohort of older women evaluated in FOOSH 1. This study also added age specific correlations of the strength and biomechanical measures from Studies 1 and 3. The strongest positive correlation was between concentric arm strength and energy absorption in both age groups, suggesting this type of contraction is important to include in strength training fall injury prevention programs.

CHAPTER 6: GENERAL DISCUSSION AND CONCLUSIONS

The objective of this dissertation was to examine the age differences in biomechanical variables, neuromuscular control and strength during a controlled and unexpected fall on the outstretched hands. Injurious falls have a range of physical consequences as falls remain the leading cause of injury- related hospitalization in Canadian seniors. By 2031, Canada's population of adults over 65 will increase from 13% to 24% [8]. This shift in demographics is also predicted to have a direct fall related healthcare cost increase from $ 2 billion to $ 4.4 billion [271]. In Canada, as our population ages, the prevention of falls has been the focus of government agencies. Although there are several studies identifying the risk factors that predict a risk of falling, there are currently no clinical guidelines that identify the modifiable factors (other than bone strength) that could decrease the risk of injury in the event when a fall in inevitable. There are some studies investigating the biomechanics and neuromuscular factors associated with simulated falls, but these are primarily in younger adults or older men. This thesis fills a gap in the literature by providing data specific to women. Data from this thesis provide evidence that both muscle strength and biomechanical factors distinguish older from younger women during a simulated FOOSH. Information from this thesis can help provide evidence to clinicians to develop targeted fall injury prevention programs.

The three studies encompassed in this thesis were designed to logically investigate specific phases of a forward FOOSH from a controlled situation to an unexpected situation. The strength of the first study is that it was designed to isolate and examine the post-impact phase of a forward fall. By simulating the weight bearing demands on the UE, this study identified important age differences in neuromuscular activity using a safe protocol with voluntary initiation of movement. It also incorporated both muscle performance and muscle activation measurement, unlike one previous study with a similar design [29]. In summary, important muscle activation and biomechanical differences were observed between younger and older women during a controlled descent on the outstretched hands despite no significant difference in UE muscle strength. Older women exhibited decreased energy absorption capabilities in their descent strategy. This strategy demonstrated decreased EO activation to stabilize the spine and created stability at the elbow by co-activating both triceps and biceps. The overall result was a decreased ability to lower their bodies as far as the younger women did.

The strength of the second study is the addition of modified components of the pre-impact and impact phases, allowing for examination of the neuromuscular contributions during each progressive phase of a FOOSH. This study describes the muscle activation patterns utilized by the young women to arrest the body during an unexpected simulated FOOSH. Deficits in the neuromuscular activation patterns prior to and after impact, when landing on the UE, could potentially increase injury risk during a FOOSH. The older women did not activate the deep core muscles: TrA and IO, in the same anticipatory manner as younger women did prior to descent, nor did they have the same pattern of UE muscle (TRI, AntDEL, PEC, BB) activation after impact.

The strength of the third study was the inclusion of biomechanical variables in which differences were expected to increase injury risk in the older women compared to the younger women. In this study, despite having no differences in muscle activity and resultant elbow stiffness, the older women absorbed far less energy with their UEs. Absorption of energy is the primary factor associated with dissipating the energy to decrease the force of the impact and risk of fracture. Fall arrest strategies were quantifiably different between age groups. The younger women demonstrated an increased elbow angular velocity and flexion angle at impact. This strategy has been cited by other laboratory studies of younger adults, as well as through computer simulation models, as the optimal way to use the UE to absorb the energy of a FOOSH.

A challenge of the study was the difficulty of simulating a natural forward fall in a controlled laboratory setting. Body positioning and falling distance were controlled for safety and, therefore, did not mimic the demands a forward fall from standing height in terms of fall height, velocity or impact forces. Having the participants start with a flexed shoulder prior to an unexpected released was necessary for safety reasons, but, unfortunately, this does not reproduce the demands of having to get the hands in position prior to impact. Although the current study did not test reactive arm positioning, it still gives insight into age-related changes in muscle recruitment. Another limitation of this thesis, although necessary to avoid excessive fatigue and test burden, was that strength and muscle activation measurement were completed in the non-dominant arm only. As well, although the novel strength testing protocol was developed to replicate a multi-joint movement similar to that required to arrest a forward fall, it was not able to account for trunk strength and stability because the testing was done in sitting.

Future research is needed to confirm age-related UE strength differences during functionally specific strength tests including tests of muscular power. Trunk muscle strength testing is also warranted based on the muscle activity differences observed in these studies. Measurement of muscle strength and muscle activation was limited to the non-dominant UE. Observing both limbs could address possible bilateral differences and address the possibility of age-related bilateral deficit. Since trunk muscle activation differences were observed in this thesis and since this activity is assumed to contribute to the stability of the spine, further investigation should involve a biomechanical analysis of the proximal link in the kinetic chain (pelvis and spine). As energy absorption was a consistent age-related difference in these studies, further work should examine which factors best predict energy absorption and what training interventions would best improve UE energy absorption capacity. Future studies should begin to investigate the effect of interventions focusing on UE and trunk muscle strength training along with specific forward fall training focusing on absorbing the energy of a forward fall.

The data presented in this thesis characterize age differences in the ability to control and arrest the forward momentum of the body during a simulated FOOSH. When the protective use of the hands is executed in this common forward fall simulation, older women land with the elbow joint in more extension than younger women and moving with lower velocity. Data from a controlled descent supports this finding, as older women also had a higher mean co-activity in their biceps/triceps during the descent and do not activate their trunk muscles comparatively to the younger women. The decreased energy absorbing capacity of the UE could increase older women's risk of injury during a FOOSH. The aforementioned differences could translate into the contributing factors behind the common differential variable of energy absorption of the UE, but is not entirely clear and therefore further research is warranted.

6.1 Conclusion

The data presented in this thesis characterize age differences in the ability to control and arrest the forward momentum of the body during a simulated FOOSH. When the protective use of the hands is executed in this common forward fall simulation, older women land with the elbow joint in more extension than younger women and move with lower velocity. Data from the controlled descent support this finding, as older women also had higher mean co-activity in their biceps/triceps during the descent and do not activate their trunk muscles comparatively to the younger women. The decreased energy absorbing capacity of the UE could increase older women's risk of injury during a FOOSH. The aforementioned differences could translate into the contributing factors behind the common differential variable of energy absorption of the UE, but is not entirely clear and therefore further research is warranted

6.2 Clinical implications

Evidence supports that exercise is an effective intervention to prevent falls [272-274]. Strength training has been defined as a key element of fall prevention programs and is more effective when combined with balance training [272-274]. The key components of strength training interventions for balance and falls reduction are: (1) strengthen lower-extremity and postural muscles, (2) perform exercises with minimal upper-extremity support, and (3) exercise at moderate or high intensity [272 273,274]. There has been no investigation of what type of training program would prevent fall related injuries when a fall is inevitable. Fall prevention programs and associated clinical practice guidelines have neglected to include recommendations for UE strengthening. There are no interventions or guidelines for "training" fall arrest strategies.

The older women who participated in these studies were healthy community dwelling adults over the age of 60 years. They reported varying levels of physical activity, but most were moderately to highly active. In fact, there were no significant differences in activity level compared to the younger cohort. This makes it difficult to apply the findings to women more at risk of fall-related injury; however, even this healthy population demonstrated differences that put them at increased risk of injury.

The healthy older women in these studies demonstrated different trunk muscle activation timing and magnitude compared to the younger women, suggesting decreased control and strength of the proximal musculature. Although it is not entirely clear how the trunk musculature contributes to the successful arrest of a forward fall, based on the high activation amplitudes displayed by the young women for TrA/IO in the pre impact phase in Study 2 and for EO during the post-impact descent phase, it can be surmised that the trunk muscles are involved in a successful forward fall descent and their relationship to injury prevention should be investigated further.

The older women landed with little modulation of UE posture, specifically with more extended elbows, which is known to be a non-optimal landing strategy. This elbow strategy could be an important target for training programs; to focus on re-learning how to control a landing with slightly flexed elbows. Neuromuscular training is often used in rehabilitation and training of athletes with sports injuries. Neuromuscular training can improve joint position sense, stability and protective reflexes [275], which could improve specific aspects of a FOOSH arrest such as increasing elbow flexion, actively "catching" the landing, increasing trunk muscle activity and decreasing BB muscle activity. Future research should investigate whether adding neuromuscular control training for forward fall strategies improves the capacity of older women to land safely

The modulation of energy absorption capabilities, by altering elbow velocity and increasing elbow flexion angles, may be an effective injury prevention tactic. Further study is needed, but this suggests that trunk and UE neuromuscular training may be beneficial to include in a fall injury prevention program for older women. Deficits were observed in the older women's UE concentric strength (CON) (in Studies 1 &3) during a multi-joint strength test and CON was correlated with energy absorption during the descent task. Multi-joint exercises recruit several muscles at a time, including the prime movers and stabilizing muscles [276]. This type of exercise requires learning and co-ordination of specific movement patterns and it can be adapted to target specific movement patterns. An exercise that replicates movement patterns and neuromuscular demands of a FOOSH arrest should be the next step in research to be to develop and test to determine if it might be a feasible and effective intervention. The push up exercise is one such exercise that might be modified to simulate the demands of a FOOSH with good specificity; however, further prospective study is needed to evaluate its safety, feasibility and effects on injury risk.

Building of the findings of these studies, I would recommend a future randomized control trial of a specific intervention for upper extremity/ trunk muscle strengthening and fall arrest strategy training combined with an existing fall prevention program. Future studies should also examine the possible relationships of upper extremity power, since arresting a fall involves high velocity contractions. Power training has proven to be safe and effective in this population, power training in the upper extremity and trunk should be investigated.

This thesis supports future study in this area to identify the potential modifiable risk factors associated with injury risk when a fall is inevitable. These factors may be important to consider in designing and evaluating future exercise programs to address preventing injury during forward falls. Upper extremity strength training including weight bearing activities on the UE should be included in multicomponent exercise programs to prevent falls and fall injuries. An evidence based program should be developed incorporating fall arrested strategy training along with UE and trunk muscle strengthening and tested in the community. The results obtained from this thesis are clinically important, as they provide evidence of the neuromuscular and biomechanical demands facing women during a FOOSH.

REFERENCES

1. Hauer K, Lamb SE, Jorstad EC, Todd C, Becker C. Systematic review of definitions and methods of measuring falls in randomised controlled fall prevention trials. *Age Ageing.* 2006;35.
2. McComas AJ. *Skeletal muscle: form and function.* Champaign, Ill.;: Human Kinetics; 1996.
3. Enoka RM, ed *Neuromechanics of Human Movement.* 3rd ed. Champaign: Human Kinetics; 2002.
4. Basmajian JV. *Muscles alive: their functions revealed by electromyography. 4th ed.* Baltimore;: Williams & Wilkins; 1978.
5. Parachute Canada: The Cost of Injury in Canada. 2015; http://www.parachutecanada.org/downloads/research/Cost_of_Injury-2015.pdf.
6. Liu SW, Obermeyer Z, Chang Y, Shankar KN. Frequency of ED revisits and death among older adults after a fall. *The American Journal Of Emergency Medicine.* 2015;33(8):1012-1018.
7. Public Health Agency of Canada. Seniors' Falls in Canada: second report, (2014).
8. Canada PHAo. Public Health Agency of Canada. Seniors' falls in Canada: second report [Internet]. 2014; : http://www.phac-aspc.gc.ca/seniors -aines/publications/public/index-eng.php. Accessed May, 2015.
9. Wolinsky FD, Bentler SE, Liu L, et al. Recent hospitalization and the risk of hip fracture among older Americans. *The journals of gerontology. Series A, Biological sciences and medical sciences.* 2009;64(2):249-255.
10. Ioannidis G, Papaioannou A, Hopman WM, et al. Relation between fractures and mortality: results from the Canadian Multicentre Osteoporosis Study. *CMAJ : Canadian Medical Association journal = journal de l'Association medicale canadienne.* 2009;181(5):265-271.
11. Jones G, Nguyen T, Sambrook PN, Kelly PJ, Gilbert C, Eisman JA. Symptomatic fracture incidence in elderly men and women: the Dubbo Osteoporosis Epidemiology Study (DOES). *Osteoporosis International: A Journal Established As Result Of Cooperation Between The European Foundation For Osteoporosis And The National Osteoporosis Foundation Of The USA.* 1994;4(5):277-282.
12. Palvanen M, Kannus P, Parkkari J, et al. The injury mechanisms of osteoporotic upper extremity fractures among older adults: a controlled study of 287 consecutive patients and their 108 controls. *Osteoporosis International: A Journal Established As Result Of Cooperation Between The European Foundation For Osteoporosis And The National Osteoporosis Foundation Of The USA.* 2000;11(10):822-831.
13. van Staa TP, Dennison EM, Leufkens HG, Cooper C. Epidemiology of fractures in England and Wales. *Bone.* 2001;29(6):517-522.
14. Chang VC, Do MT. Risk Factors for Falls Among Seniors: Implications of Gender. *American journal of epidemiology.* 2015;181(7):521-531.
15. Stevens JA, Sogolow ED. Gender differences for non-fatal unintentional fall related injuries among older adults. *Injury Prevention: Journal Of The International Society For Child And Adolescent Injury Prevention.* 2005;11(2):115-119.
16. Thibaud M, Bloch F, Tournoux-Facon C, et al. Impact of physical activity and sedentary behaviour on fall risks in older people: a systematic review and meta-analysis of observational studies. *European Reviews of Aging & Physical Activity.* 2012;9(1):5-15.

17. Cummings SR, Nevitt MC. A hypothesis: the causes of hip fractures. *Journal Of Gerontology.* 1989;44(4):M107-M111.
18. Stel VS, Smit JH, Pluijm SMF, Lips P. Consequences of falling in older men and women and risk factors for health service use and functional decline. *Age And Ageing.* 2004;33(1):58-65.
19. Granacher U, Lacroix A, Roettger K, Gollhofer A, Muehlbauer T. Relationships Between Trunk Muscle Strength, Spinal Mobility, and Balance Performance in Older Adults. *Journal of Aging & Physical Activity.* 2014;22(4):490-498.
20. DeGoede KM, Ashton-Miller JA, Schultz AB. Fall-related upper body injuries in the older adult: a review of the biomechanical issues. *Journal of biomechanics.* 2003;36(7):1043-1053.
21. Nevitt MC, Cummings SR. Type of fall and risk of hip and wrist fractures: the study of osteoporotic fractures. The Study of Osteoporotic Fractures Research Group. *Journal of the American Geriatrics Society.* 1993;41(11):1226-1234.
22. DeGoede KM, Ashton-Miller JA. Fall arrest strategy affects peak hand impact force in a forward fall. *Journal of biomechanics.* 2002;35(6):843-848.
23. DeGoede KM, Ashton-Miller JA. Biomechanical simulations of forward fall arrests: effects of upper extremity arrest strategy, gender and aging-related declines in muscle strength. *Journal of biomechanics.* 2003;36(3):413-420.
24. Faulkner JA, Larkin LM, Claflin DR, Brooks SV. Age-related changes in the structure and function of skeletal muscles. *Clinical and experimental pharmacology & physiology.* 2007;34(11):1091-1096.
25. Berger MJ, Doherty TJ. Sarcopenia: prevalence, mechanisms, and functional consequences. *Interdisciplinary Topics In Gerontology.* 2010;37:94-114.
26. Sorensen MB, Rosenfalck AM, Hojgaard L, Ottesen B. Obesity and sarcopenia after menopause are reversed by sex hormone replacement therapy. *Obesity research.* 2001;9(10):622-626.
27. Sipila S, Taaffe DR, Cheng S, Puolakka J, Toivanen J, Suominen H. Effects of hormone replacement therapy and high-impact physical exercise on skeletal muscle in post-menopausal women: a randomized placebo-controlled study. *Clinical science.* 2001;101(2):147-157.
28. Lang T, Streeper T, Cawthon P, Baldwin K, Taaffe DR, Harris TB. Sarcopenia: etiology, clinical consequences, intervention, and assessment. *Osteoporosis International: A Journal Established As Result Of Cooperation Between The European Foundation For Osteoporosis And The National Osteoporosis Foundation Of The USA.* 2010;21(4):543-559.
29. Sran MM, Stotz PJ, Normandin SC, Robinovitch SN. Age differences in energy absorption in the upper extremity during a descent movement: implications for arresting a fall. *The journals of gerontology. Series A, Biological sciences and medical sciences.* 2010;65(3):312-317.
30. Woolrych R, Zecevic A, Sixsmith A, et al. Using Video Capture to Investigate the Causes of Falls in Long-Term Care. *Gerontologist.* 2015;55(3):483-494 412p.
31. Gillespie LD, Robertson MC, Gillespie WJ, et al. Interventions for preventing falls in older people living in the community. *The Cochrane Database Of Systematic Reviews.* 2012;9:CD007146.

32. Arnold CM, Sran MM, Harrison EL. Exercise for fall risk reduction in community-dwelling older adults: a systematic review. *Physiotherapy Canada.* 2008;60(4):358-372.
33. Shubert TE. Evidence-based exercise prescription for balance and falls prevention: a current review of the literature. *Journal Of Geriatric Physical Therapy (2001).* 2011;34(3):100-108.
34. Trudelle-Jackson EJ, Jackson AW, Morrow Jr JR. Muscle Strength and Postural Stability in Healthy, Older Women: Implications for Fall Prevention. *Journal of Physical Activity & Health.* 2006;3(3):292-303.
35. Tinetti ME, Speechley M, Ginter SF. Risk factors for falls among elderly persons living in the community. *The New England journal of medicine.* 1988;319.
36. Tromp AM, Pluijm SMF, Smit JH, Deeg DJH, Bouter LM, Lips P. Fall-risk screening test: a prospective study on predictors for falls in community-dwelling elderly. *J Clin Epidemiol.* 2001;54.
37. Nevitt MC, Cummings SR, Hudes ES. Risk factors for injurious falls: a prospective study. *Journal Of Gerontology.* 1991;46(5):M164-M170.
38. Kannus P, Sievänen H, Palvanen M, Järvinen T, Parkkari J. Prevention of falls and consequent injuries in elderly people. *Lancet (London, England).* 2005;366(9500):1885-1893.
39. Karlsson MK, Magnusson H, von Schewelov T, Rosengren BE. Prevention of falls in the elderly--a review. *Osteoporosis International: A Journal Established As Result Of Cooperation Between The European Foundation For Osteoporosis And The National Osteoporosis Foundation Of The USA.* 2013;24(3):747-762.
40. Chou PH, Chou YL, Lin CJ, et al. Effect of elbow flexion on upper extremity impact forces during a fall. *Clinical Biomechanics (Bristol, Avon).* 2001;16(10):888-894.
41. Kim KJ, Ashton-Miller JA. Biomechanics of fall arrest using the upper extremity: age differences. *Clinical biomechanics.* 2003;18(4):311-318.
42. Lo J, McCabe GN, DeGoede KM, Okuizumi H, Ashton-Miller JA. On reducing hand impact force in forward falls: results of a brief intervention in young males. *Clinical biomechanics.* 2003;18(8):730-736.
43. Hauer K, Becker C, Lindemann U, Beyer N. Effectiveness of physical training on motor performance and fall prevention in cognitively impaired older persons: a systematic review. *American Journal Of Physical Medicine & Rehabilitation / Association Of Academic Physiatrists.* 2006;85(10):847-857.
44. Woolcott JC, Khan KM, Mitrovic S, Anis AH, Marra CA. The cost of fall related presentations to the ED: a prospective, in-person, patient-tracking analysis of health resource utilization. *Osteoporosis international : a journal established as result of cooperation between the European Foundation for Osteoporosis and the National Osteoporosis Foundation of the USA.* 2012;23(5):1513-1519.
45. Rivara FP, Grossman DC, Cummings P. Injury prevention. First of two parts. *The New England journal of medicine.* 1997;337(8):543-548.
46. Giangregorio LM, MacIntyre NJ, Heinonen A, et al. Too Fit To Fracture: a consensus on future research priorities in osteoporosis and exercise. *Osteoporosis International: A Journal Established As Result Of Cooperation Between The European Foundation For Osteoporosis And The National Osteoporosis Foundation Of The USA.* 2014;25(5):1465-1472.

47. O'Neill TW, Varlow J, Silman AJ, et al. Age and sex influences on fall characteristics. *Annals of the rheumatic diseases.* 1994;53(11):773-775.
48. Schonnop R, Yang Y, Feldman F, Robinson E, Loughin M, Robinovitch SN. Prevalence of and factors associated with head impact during falls in older adults in long-term care. *CMAJ : Canadian Medical Association journal = journal de l'Association medicale canadienne.* 2013;185(17):E803-810.
49. Yang Y, Mackey DC, Liu-Ambrose T, Feldman F, Robinovitch SN. Risk factors for hip impact during real-life falls captured on video in long-term care. *Osteoporosis International: A Journal Established As Result Of Cooperation Between The European Foundation For Osteoporosis And The National Osteoporosis Foundation Of The USA.* 2016;27(2):537-547.
50. Hsiao ET, Robinovitch SN. Common protective movements govern unexpected falls from standing height. *Journal of biomechanics.* 1998;31(1):1-9.
51. Feldman F, Robinovitch SN. Reducing hip fracture risk during sideways falls: evidence in young adults of the protective effects of impact to the hands and stepping. *Journal of biomechanics.* 2007;40(12):2612-2618.
52. Chiu J, Robinovitch SN. Prediction of upper extremity impact forces during falls on the outstretched hand. *Journal of biomechanics.* 1998;31(12):1169-1176.
53. Augat P, Iida H, Jiang Y, Diao E, Genant HK. Distal radius fractures: mechanisms of injury and strength prediction by bone mineral assessment. *Journal Of Orthopaedic Research: Official Publication Of The Orthopaedic Research Society.* 1998;16(5):629-635.
54. Santello M. Review of motor control mechanisms underlying impact absorption from falls. *Gait & Posture.* 2005;21(1):85-94.
55. Dietz V, Noth J, Schmidtbleicher D. Interaction between pre-activity and stretch reflex in human triceps brachii during landing from forward falls. *The Journal Of Physiology.* 1981;311:113-125.
56. Lo J, Ashton-Miller JA. Effect of upper and lower extremity control strategies on predicted injury risk during simulated forward falls: a study in healthy young adults. *Journal Of Biomechanical Engineering.* 2008;130(4):041015-041015.
57. DeGoede KM, Ashton-Miller JA, Schultz AB, Alexander NB. Biomechanical factors affecting the peak hand reaction force during the bimanual arrest of a moving mass. *J Biomech Eng.* 2002;124(1):107-112.
58. Sattin RW, Lambert Huber DA, DeVito CA, et al. The incidence of fall injury events among the elderly in a defined population. *American journal of epidemiology.* 1990;131(6):1028-1037.
59. Riggs BL, Melton LJ, 3rd, Robb RA, et al. Population-based analysis of the relationship of whole bone strength indices and fall-related loads to age- and sex-specific patterns of hip and wrist fractures. *Journal Of Bone And Mineral Research: The Official Journal Of The American Society For Bone And Mineral Research.* 2006;21(2):315-323.
60. Vogt MT, Cauley JA, Tomaino MM, Stone K, Williams JR, Herndon JH. Distal Radius Fractures in Older Women: A 10-Year Follow-Up Study of Descriptive Characteristics and Risk Factors. The Study of Osteoporotic Fractures. *Journal of the American Geriatrics Society.* 2002;50(1):97-103.

61. Yang Y, Feldman F, Leung PM, Scott V, Robinovitch SN. Agreement Between Video Footage and Fall Incident Reports on the Circumstances of Falls in Long-Term Care. *Journal of the American Medical Directors Association.* 2015;16(5):388-394 387p.
62. Robinovitch, Feldman F, Yang Y, et al. Video capture of the circumstances of falls in elderly people residing in long-term care: an observational study. *Lancet.* 2013;381 North American Edition(9860):47-54 48p.
63. Choi WJ, Wakeling JM, Robinovitch SN. Kinematic analysis of video-captured falls experienced by older adults in long-term care. *Journal of biomechanics.* 2015;48(6):911-920.
64. DeGoede KM, Ashton-Miller JA, Liao JM, Alexander NB. How quickly can healthy adults move their hands to intercept an approaching object? Age and gender effects. *The journals of gerontology. Series A, Biological sciences and medical sciences.* 2001;56(9):M584-588.
65. Robinovitch SN, Chiu J. Surface stiffness affects impact force during a fall on the outstretched hand. *Journal Of Orthopaedic Research: Official Publication Of The Orthopaedic Research Society.* 1998;16(3):309-313.
66. Magnus CRA, Arnold CM, Johnston G, et al. Cross-Education for Improving Strength and Mobility After Distal Radius Fractures: A Randomized Controlled Trial. *Archives of Physical Medicine & Rehabilitation.* 2013;94(7):1247-1255 1249p.
67. Arnold CD-B-HV, Farthing J, K. Crockett, C. Haver, G. Johnston, J. Basran. Factors contributing to falls resulting in wrist fracture and the relationship to functional status in women over the age of 50 years. *Canadian Journal on Aging.* 2015.
68. Arnold CM, Busch AJ, Schachter CL, Harrison L, Olszynski W. The relationship of intrinsic fall risk factors to a recent history of falling in older women with osteoporosis. *The Journal Of Orthopaedic And Sports Physical Therapy.* 2005;35(7):452-460.
69. Berg K. Balance and its measure in the elderly: a review. *Physiotherapy Canada.* 1989;41(5):240-246 247p.
70. Grabiner MD, Donovan S, Bareither ML, et al. Trunk kinematics and fall risk of older adults: translating biomechanical results to the clinic. *Journal of electromyography and kinesiology : official journal of the International Society of Electrophysiological Kinesiology.* 2008;18(2):197-204.
71. Horak FB. Postural orientation and equilibrium: what do we need to know about neural control of balance to prevent falls? *Age & Ageing.* 2006;35:ii7-ii11 11p.
72. Aruin ASS. Anticipatory postural adjustments in conditions of postural instability. *Electroencephalography and clinical neurophysiology. Electromyography and motor control.* 1998;109(4):350-359.
73. Bouisset S, Zattara M. Biomechanical study of the programming of anticipatory postural adjustments associated with voluntary movement. *Journal of biomechanics.* 1987;20(8):735-742.
74. Tang PF, Woollacott MH, Chong RK. Control of reactive balance adjustments in perturbed human walking: roles of proximal and distal postural muscle activity. *Experimental Brain Research.* 1998;119(2):141-152.
75. Cruz-Jentoft AJ, Baeyens JP, Bauer JM, et al. Sarcopenia: European consensus on definition and diagnosis: Report of the European Working Group on Sarcopenia in Older People. *Age And Ageing.* 2010;39(4):412-423.

76. Goodpaster BH, Park SW, Harris TB, et al. The loss of skeletal muscle strength, mass, and quality in older adults: the health, aging and body composition study. *The journals of gerontology. Series A, Biological sciences and medical sciences.* 2006;61(10):1059-1064.
77. Hughes VA, Frontera WR, Wood M, et al. Longitudinal Muscle Strength Changes in Older Adults: Influence of Muscle Mass, Physical Activity, and Health. *Journals of Gerontology Series A: Biological Sciences & Medical Sciences.* 2001;56A(5):B209-B217.
78. Beliaeff S, Bouchard DR, Hautier C, Brochu M, Dionne IJ. Association between muscle mass and isometric muscle strength in well-functioning older men and women. *Journal of Aging & Physical Activity.* 2008;16(4):484-493 410p.
79. Clark BC, Manini TM. Sarcopenia ≠ Dynapenia. *Journals of Gerontology Series A: Biological Sciences & Medical Sciences.* 2008;63A(8):829-834.
80. Visser M, Goodpaster BH, Kritchevsky SB, et al. Muscle mass, muscle strength, and muscle fat infiltration as predictors of incident mobility limitations in well-functioning older persons. *The journals of gerontology. Series A, Biological sciences and medical sciences.* 2005;60(3):324-333.
81. Delmonico MJ, Harris TB, Visser M, et al. Longitudinal study of muscle strength, quality, and adipose tissue infiltration. *The American Journal Of Clinical Nutrition.* 2009;90(6):1579-1585.
82. Narici MV, Maffulli N, Maganaris CN. Ageing of human muscles and tendons. *Disability & Rehabilitation.* 2008;30(20-22):1548-1554 1547p.
83. D'Antona G, Pellegrino MA, Adami R, et al. The effect of ageing and immobilization on structure and function of human skeletal muscle fibres. *J Physiol.* 2003;552(Pt 2):499-511.
84. Visser M, Schaap LA. Consequences of sarcopenia. *Clinics In Geriatric Medicine.* 2011;27(3):387-399.
85. Visser M, Deeg DJ, Lips P, Harris TB, Bouter LM. Skeletal muscle mass and muscle strength in relation to lower-extremity performance in older men and women. *Journal of the American Geriatrics Society.* 2000;48(4):381-386.
86. Newman AB, Kupelian V, Visser M, et al. Strength, but not muscle mass, is associated with mortality in the health, aging and body composition study cohort. *The journals of gerontology. Series A, Biological sciences and medical sciences.* 2006;61(1):72-77.
87. Manini TM, Clark BC. Dynapenia and aging: an update. *Journals of Gerontology Series A: Biological Sciences & Medical Sciences.* 2012;67A(1):28-40 13p.
88. Sayer AA, Syddall HE, Martin HJ, Dennison EM, Anderson FH, Cooper C. Falls, sarcopenia, and growth in early life: findings from the Hertfordshire Cohort Study. *American journal of epidemiology.* 2006;164(7):665-671 667p.
89. Furrer R, van Schoor NM, de Haan A, Lips P, de Jongh RT. Gender-specific associations between physical functioning, bone quality, and fracture risk in older people. *Calcified Tissue International.* 2014;94(5):522-530.
90. Scott D, Stuart AL, Kay D, Ebeling PR, Nicholson G, Sanders KM. Investigating the predictive ability of gait speed and quadriceps strength for incident falls in community-dwelling older women at high risk of fracture. *Archives Of Gerontology And Geriatrics.* 2014;58(3):308-313.
91. Cummings SR, Nevitt MC, Browner WS, et al. Risk factors for hip fracture in white women. Study of Osteoporotic Fractures Research Group. *The New England journal of medicine.* 1995;332(12):767-773.

92. Roig M, MacIntyre DL, Eng JJ, Narici MV, Maganaris CN, Reid WD. Preservation of eccentric strength in older adults: Evidence, mechanisms and implications for training and rehabilitation. *Experimental Gerontology.* 2010;45(6):400-409.
93. Vandervoort AA. Aging of the human neuromuscular system. *Muscle & Nerve.* 2002;25(1):17-25.
94. Hortobagyi T, Zheng D, Weidner M, Lambert NJ, Westbrook S, Houmard JA. The influence of aging on muscle strength and muscle fiber characteristics with special reference to eccentric strength. *The journals of gerontology. Series A, Biological sciences and medical sciences.* 1995;50(6):B399-406.
95. Hortobagyi T, De Vita P. Favorable Neuromuscular and Cardiovascular Responses to 7 Days of Exercise With an Eccentric Overload in Elderly Women. *Journals of Gerontology Series A: Biological Sciences & Medical Sciences.* 2000;55A(8):B401-B410.
96. Poulin MJ, Vandervoort AA, Paterson DH, Kramer JF, Cunningham DA. Eccentric and concentric torques of knee and elbow extension in young and older men. *Canadian Journal Of Sport Sciences = Journal Canadien Des Sciences Du Sport.* 1992;17(1):3-7.
97. Hortobágyi T, Devita P. Mechanisms responsible for the age-associated increase in coactivation of antagonist muscles. *Exercise And Sport Sciences Reviews.* 2006;34(1):29-35.
98. Power GA, Makrakos DP, Stevens DE, Rice CL, Vandervoort AA. Velocity dependence of eccentric strength in young and old men: the need for speed! *Applied Physiology, Nutrition, And Metabolism = Physiologie Appliquée, Nutrition Et Métabolisme.* 2015;40(7):703-710.
99. Foldvari M, Clark M, Laviolette LC, et al. Association of muscle power with functional status in community-dwelling elderly women. *The journals of gerontology. Series A, Biological sciences and medical sciences.* 2000;55(4):M192-M199.
100. Bean JF, Kiely DK, LaRose S, Goldstein R, Frontera WR, Leveille SG. Are changes in leg power responsible for clinically meaningful improvements in mobility in older adults? *Journal of the American Geriatrics Society.* 2010;58(12):2363-2368 2366p.
101. Bean JF, Leveille SG, Kiely DK, Bandinelli S, Guralnik JM, Ferrucci L. A comparison of leg power and leg strength within the InCHIANTI study: which influences mobility more? *The journals of gerontology. Series A, Biological sciences and medical sciences.* 2003;58(8):728-733.
102. Boncompagni S, d'Amelio L, Fulle S, Fanò G, Protasi F. Progressive disorganization of the excitation-contraction coupling apparatus in aging human skeletal muscle as revealed by electron microscopy: a possible role in the decline of muscle performance. *The journals of gerontology. Series A, Biological sciences and medical sciences.* 2006;61(10):995-1008.
103. Roos MR, Rice CL, Connelly DM, Vandervoort AA. Quadriceps muscle strength, contractile properties, and motor unit firing rates in young and old men. *Muscle & Nerve.* 1999;22(8):1094-1103.
104. Amaral JF, Alvim FC, Castro EA, Doimo LA, Silva MV, Novo Júnior JM. Influence of aging on isometric muscle strength, fat-free mass and electromyographic signal power of the upper and lower limbs in women. *Brazilian Journal Of Physical Therapy.* 2014;18(2):183-190.
105. Hortobagyi T, DeVita P. Muscle pre- and coactivity during downward stepping are associated with leg stiffness in aging. *Journal of electromyography and kinesiology :*

official journal of the International Society of Electrophysiological Kinesiology. 2000;10(2):117-126.
106. Nagai K, Yamada M, Uemura K, Yamada Y, Ichihashi N, Tsuboyama T. Differences in muscle coactivation during postural control between healthy older and young adults. *Archives Of Gerontology And Geriatrics.* 2011;53(3):338-343.
107. Mian OS, Thom JM, Ardigò LP, Narici MV, Minetti AE. Metabolic cost, mechanical work, and efficiency during walking in young and older men. *Acta Physiologica (Oxford, England).* 2006;186(2):127-139.
108. Schmitz A, Silder A, Heiderscheit B, Mahoney J, Thelen DG. Differences in lower-extremity muscular activation during walking between healthy older and young adults. *Journal Of Electromyography And Kinesiology: Official Journal Of The International Society Of Electrophysiological Kinesiology.* 2009;19(6):1085-1091.
109. Barry BK, Warman GE, Carson RG. Age-related differences in rapid muscle activation after rate of force development training of the elbow flexors. *Experimental Brain Research.* 2005;162(1):122-132.
110. Laroche DP, Knight CA, Dickie JL, Lussier M, Roy SJ. Explosive force and fractionated reaction time in elderly low- and high-active women. *Medicine & Science in Sports & Exercise.* 2007;39(9):1659-1665 1657p.
111. Klass M, Baudry S, Duchateau J. Age-related decline in rate of torque development is accompanied by lower maximal motor unit discharge frequency during fast contractions. *Journal of applied physiology.* 2008;104(3):739-746.
112. Bean JF, Kiely DK, Herman S, et al. The relationship between leg power and physical performance in mobility-limited older people. *Journal of the American Geriatrics Society.* 2002;50(3):461-467 467p.
113. Clark BC, Manini TM. Functional consequences of sarcopenia and dynapenia in the elderly. *Current Opinion in Clinical Nutrition & Metabolic Care.* 2010;13(3):271-276 276p.
114. Patten C, Kamen G, Rowland DM. Adaptations in maximal motor unit discharge rate to strength training in young and older adults. *Muscle & Nerve.* 2001;24(4):542-550.
115. Leong B, Kamen G, Patten C, Burke JR. Maximal motor unit discharge rates in the quadriceps muscles of older weight lifters. / Frequence maximale de decharge des unites motrices des muscles quadriceps d'halterophiles ages. *Medicine & Science in Sports & Exercise.* 1999;31(11):1638-1644.
116. Klitgaard H, Mantoni M, Schiaffino S, et al. Function, morphology and protein expression of ageing skeletal muscle : a cross-sectional study of elderly men with different training backgrounds (Caracteristiques fonctionnelles, morphologie et expressivite des proteines musculaires dans le muscle squelettique vieillissant : etude transversale d'hommes ages ayant des passes d'entrainement differents. *Acta Physiologica Scandinavica.* 1990;140(1):41-54.
117. Lexell J. Human aging, muscle mass, and fiber type composition. *Journals of Gerontology Series A: Biological Sciences & Medical Sciences.* 1995;50A(6):11.
118. Lynch NA, Metter EJ, Lindle RS, et al. Muscle quality. I. Age-associated differences between arm and leg muscle groups. *Journal of applied physiology.* 1999;86(1):188-194.
119. Candow DG, Chilibeck PD. Differences in size, strength, and power of upper and lower body muscle groups in young and older men. *Journals of Gerontology Series A: Biological Sciences & Medical Sciences.* 2005;60A(2):148-156 149p.

120. Hughes VA, Frontera WR, Wood M, et al. Longitudinal muscle strength changes in older adults: influence of muscle mass, physical activity, and health. *The journals of gerontology. Series A, Biological sciences and medical sciences.* 2001;56(5):B209-B217.
121. Samuel D, Wilson K, Martin HJ, Allen R, Sayer AA, Stokes M. Age-associated changes in hand grip and quadriceps muscle strength ratios in healthy adults. *Aging clinical and experimental research.* 2012;24(3):245-250.
122. Hicks GE, Shardell M, Alley DE, et al. Absolute strength and loss of strength as predictors of mobility decline in older adults: the InCHIANTI study. *The journals of gerontology. Series A, Biological sciences and medical sciences.* 2012;67(1):66-73.
123. Hairi NN, Cumming RG, Naganathan V, et al. Loss of muscle strength, mass (sarcopenia), and quality (specific force) and its relationship with functional limitation and physical disability: the Concord Health and Ageing in Men Project. *Journal of the American Geriatrics Society.* 2010;58(11):2055-2062.
124. Lauretani F, Russo CR, Bandinelli S, et al. Age-associated changes in skeletal muscles and their effect on mobility: an operational diagnosis of sarcopenia. *Journal Of Applied Physiology (Bethesda, Md.: 1985).* 2003;95(5):1851-1860.
125. Crockett K, Ardell K, Hermanson M, et al. The Relationship of Knee-Extensor Strength and Rate of Torque Development to Sit-to-Stand Performance in Older Adults. *Physiotherapy Canada.* 2013;65(3):229-235.
126. Abe T, Thiebaud RS, Loenneke JP, Ogawa M, Mitsukawa N. Association between forearm muscle thickness and age-related loss of skeletal muscle mass, handgrip and knee extension strength and walking performance in old men and women: a pilot study. *Ultrasound In Medicine & Biology.* 2014;40(9):2069-2075.
127. Taekema DG, Gussekloo J, Maier AB, Westendorp RGJ, de Craen AJM. Handgrip strength as a predictor of functional, psychological and social health. A prospective population-based study among the oldest old. *Age And Ageing.* 2010;39(3):331-337.
128. Cheung C-L, Tan KCB, Bow CH, Soong CSS, Loong CHN, Kung AW-C. Low handgrip strength is a predictor of osteoporotic fractures: cross-sectional and prospective evidence from the Hong Kong Osteoporosis Study. *Age (Dordrecht, Netherlands).* 2012;34(5):1239-1248.
129. Kojima N, Kim M, Saito K, et al. Lifestyle-Related Factors Contributing to Decline in Knee Extension Strength among Elderly Women: A Cross-Sectional and Longitudinal Cohort Study. *Plos One.* 2015;10(7):e0132523-e0132523.
130. Nakao H, Yoshikawa T, Mimura T, Hara T, Nishimoto K, Fujimoto S. Influence of Lower-extremity Muscle Force, Muscle Mass and Asymmetry in Knee Extension Force on Gait Ability in Community-dwelling Elderly Women. *Journal of Physical Therapy Science.* 2006;18(1):73-79.
131. Yamada T, Demura S. The relationship of force output characteristics during a sit-to-stand movement with lower limb muscle mass and knee joint extension in the elderly. *Archives Of Gerontology And Geriatrics.* 2010;50(3):e46-e50.
132. Martien S, Delecluse C, Boen F, et al. Is knee extension strength a better predictor of functional performance than handgrip strength among older adults in three different settings? *Archives of Gerontology & Geriatrics.* 2015;60(2):252-258.
133. Metter EJ, Talbot LA, Schrager M, Conwit R. Skeletal muscle strength as a predictor of all-cause mortality in healthy men. *The journals of gerontology. Series A, Biological sciences and medical sciences.* 2002;57(10):B359-B365.

134. Kallman DA, Plato CC, Tobin JD. The role of muscle loss in the age-related decline of grip strength: cross-sectional and longitudinal perspectives. *Journal Of Gerontology*. 1990;45(3):M82-M88.
135. Wang C, Chen L. Grip Strength in Older Adults: Test-Retest Reliability and Cutoff for Subjective Weakness of Using the Hands in Heavy Tasks. *Archives of Physical Medicine & Rehabilitation.* 2010;91(11):1747-1751 1745p.
136. Tietjen-Smith T, Smith SW, Martin M, Henry R, Weeks S, Bryant A. Grip strength in relation to overall strength and functional capacity in very old and oldest old females. *Physical and Occupational Therapy in Geriatrics.* 2006;24(4):63-78.
137. Pentland WE, Vandervoort AA, Twomey LT. Age-related changes in upper limb isokinetic and grip strength. *Physiotherapy Theory & Practice.* 1995;11(3):165-173 169p.
138. Ikeda N, Murata S, Otao H, Murata JUN, Horie JUN, Mizota K. The Relationship between Grip Strength and Physical Function in Elderly Community-Dwelling Women. *Rigakuryoho Kagaku.* 2011;26(2):255-258.
139. Rikli RA, Jones CJ. Assessing Physical Performance in Independent Older Adults: Issues and Guidelines. *Journal of Aging & Physical Activity.* 1997;5(3):244.
140. Landers KA, Hunter GR, Wetzstein CJ, Bamman MM, Weinsier RL. The interrelationship among muscle mass, strength, and the ability to perform physical tasks of daily living in younger and older women. *The journals of gerontology. Series A, Biological sciences and medical sciences.* 2001;56(10):B443-B448.
141. Runnels ED, Bemben DA, Anderson MA, Bemben MG. Influence of age on isometric, isotonic, and isokinetic force production characteristics in men. *Journal Of Geriatric Physical Therapy (2001).* 2005;28(3):74-84.
142. McGee CW, Mathiowetz V. The relationship between upper extremity strength and instrumental activities of daily living performance among elderly women. *OTJR: Occupation, Participation & Health.* 2003;23(4):143-154 112p.
143. Clark DJ, Pojednic RM, Reid KF, et al. Longitudinal Decline of Neuromuscular Activation and Power in Healthy Older Adults. *Journals of Gerontology Series A: Biological Sciences & Medical Sciences.* 2013;68(11):1419-1425.
144. Drouin JM, Valovich-mcLeod TC, Shultz SJ, Gansneder BM, Perrin DH. Reliability and validity of the Biodex system 3 pro isokinetic dynamometer velocity, torque and position measurements. *European Journal Of Applied Physiology.* 2004;91(1):22-29.
145. Clark DJ, Fielding RA. Neuromuscular contributions to age-related weakness. *The journals of gerontology. Series A, Biological sciences and medical sciences.* 2012;67(1):41-47.
146. Milner-Brown HS, Stein RB. The relation between the surface electromyogram and muscular force. *The Journal Of Physiology.* 1975;246(3):549-569.
147. Moritani T, deVries HA. Reexamination of the relationship between the surface integrated electromyogram (IEMG) and force of isometric contraction. *American Journal Of Physical Medicine.* 1978;57(6):263-277.
148. Narici MV, Bordini M, Cerretelli P. Effect of aging on human adductor pollicis muscle function. *Journal of applied physiology.* 1991;71(4):1277-1281.
149. Frontera WR, Suh D, Krivickas LS, Hughes VA, Goldstein R, Roubenoff R. Skeletal muscle fiber quality in older men and women. *American Journal Of Physiology. Cell Physiology.* 2000;279(3):C611-C618.

150. Miljkovic N, Lim J-Y, Miljkovic I, Frontera WR. Aging of skeletal muscle fibers. *Annals Of Rehabilitation Medicine.* 2015;39(2):155-162.
151. Kostek MC, Delmonico MJ. Age-related changes in adult muscle morphology. *Current Aging Science.* 2011;4(3):221-233.
152. Bassey EJ, Short AH. A new method for measuring power output in a single leg extension: feasibility, reliability and validity. *European Journal Of Applied Physiology And Occupational Physiology.* 1990;60(5):385-390.
153. Massion J. Movement, posture and equilibrium: interaction and coordination. *Progress In Neurobiology.* 1992;38(1):35-56.
154. Santos MJ, Kanekar N, Aruin AS. The role of anticipatory postural adjustments in compensatory control of posture: 1. Electromyographic analysis. *Journal Of Electromyography And Kinesiology: Official Journal Of The International Society Of Electrophysiological Kinesiology.* 2010;20(3):388-397.
155. Bugnariu N, Sveistrup H. Age-related changes in postural responses to externally- and self-triggered continuous perturbations. *Archives Of Gerontology And Geriatrics.* 2006;42(1):73-89.
156. Nashner LM, Cordo PJ. Relation of automatic postural responses and reaction-time voluntary movements of human leg muscles. *Experimental Brain Research.* 1981;43(3-4):395-405.
157. Burleigh AL, Horak FB, Malouin F. Modification of postural responses and step initiation: evidence for goal-directed postural interactions. *Journal Of Neurophysiology.* 1994;72(6):2892-2902.
158. Trivedi H, Leonard JA, Ting LH, Stapley PJ. Postural responses to unexpected perturbations of balance during reaching. *Experimental Brain Research.* 2010;202(2):485-491.
159. Enoka RM. *Neuromechanics of human movement. 3rd ed.* Champaign, Ill.;: Human Kinetics; 2002.
160. Fu SN, Hui-Chan CWY. Mental set can modulate response onset in the lower limb muscles to falls in humans. *Neuroscience Letters.* 2002;321(1-2):77-80.
161. McKinley PA, Smith JL, Gregor RJ. Responses of elbow extensors to landing forces during jump downs in cats. *Experimental Brain Research.* 1983;49(2):218-228.
162. Robinovitch SN, Normandin SC, Stotz P, Maurer JD. Time requirement for young and elderly women to move into a position for breaking a fall with outstretched hands. *The journals of gerontology. Series A, Biological sciences and medical sciences.* 2005;60(12):1553-1557.
163. McIlroy WE, Maki BE. Early activation of arm muscles follows external perturbation of upright stance. *Neuroscience Letters.* 1995;184(3):177-180.
164. Maki BE, McIlroy WE. Control of rapid limb movements for balance recovery: age-related changes and implications for fall prevention. *Age & Ageing.* 2006;35:ii12-ii18 11p.
165. Ochala J, Frontera WR, Dorer DJ, Van Hoecke J, Krivickas LS. Single skeletal muscle fiber elastic and contractile characteristics in young and older men. *The journals of gerontology. Series A, Biological sciences and medical sciences.* 2007;62(4):375-381.
166. Baudry S, Klass M, Pasquet B, Duchateau J. Age-related fatigability of the ankle dorsiflexor muscles during concentric and eccentric contractions. *European Journal of Applied Physiology.* 2007;100(5):515-525.

167. Lee Y, Ashton-Miller JA. Effects of Age, Gender and Level of Co-contraction on Elbow and Shoulder Rotational Stiffness and Damping in the Impulsively End-Loaded Upper Extremity. *Annals Of Biomedical Engineering.* 2015;43(5):1112-1122.
168. Baudry S, Lecoeuvre G, Duchateau J. Age-related changes in the behavior of the muscle-tendon unit of the gastrocnemius medialis during upright stance. *Journal Of Applied Physiology (Bethesda, Md.: 1985).* 2012;112(2):296-304.
169. Taube W, Leukel C, Schubert M, Gruber M, Rantalainen T, Gollhofer A. Differential modulation of spinal and corticospinal excitability during drop jumps. *J Neurophysiol.* 2008;99(3):1243-1252.
170. Blackburn JT, Padua DA, Guskiewicz KM. Muscle stiffness and spinal stretch reflex sensitivity in the triceps surae. *J Athl Train.* 2008;43(1):29-36.
171. van der Burg JCE, Pijnappels M, van Dieën JH. The influence of artificially increased trunk stiffness on the balance recovery after a trip. *Gait & Posture.* 2007;26(2):272-278.
172. Butler RJ, Crowell HP, 3rd, Davis IM. Lower extremity stiffness: implications for performance and injury. *Clinical Biomechanics (Bristol, Avon).* 2003;18(6):511-517.
173. Freeman S, Karpowicz A, Gray J, McGill S. Quantifying muscle patterns and spine load during various forms of the push-up. *Medicine and science in sports and exercise.* 2006;38(3):570-577.
174. Juker D, McGill S, Kropf P, Steffen T. Quantitative intramuscular myoelectric activity of lumbar portions of psoas and the abdominal wall during a wide variety of tasks. *Medicine and science in sports and exercise.* 1998;30(2):301-310.
175. Willson JD, Dougherty CP, Ireland ML, Davis IM. Core stability and its relationship to lower extremity function and injury. *The Journal Of The American Academy Of Orthopaedic Surgeons.* 2005;13(5):316-325.
176. Hodges PW, Richardson CA. Relationship between limb movement speed and associated contraction of the trunk muscles. *Ergonomics.* 1997;40(11):1220-1230.
177. Behm DG, Drinkwater EJ, Willardson JM, Cowley PM. The use of instability to train the core musculature. *Appl Physiol Nutr Metab.* 2010;35(1):91-108.
178. Kibler WB, Press J, Sciascia A. The role of core stability in athletic function. *Sports Med.* 2006;36(3):189-198.
179. Allum JHJ, Carpenter MG, Honegger F, Adkin AL, Bloem BR. Age-dependent variations in the directional sensitivity of balance corrections and compensatory arm movements in man. *The Journal Of Physiology.* 2002;542(Pt 2):643-663.
180. Smith MD, Coppieters MW, Hodges PW. Is balance different in women with and without stress urinary incontinence? *Neurourology and Urodynamics.* 2008;27(1):71-78.
181. Hodges PW, Cresswell AG, Daggfeldt K, Thorstensson A. Three dimensional preparatory trunk motion precedes asymmetrical upper limb movement. *Gait Posture.* 2000;11(2):92-101.
182. Panjabi MM. The stabilizing system of the spine. Part II. Neutral zone and instability hypothesis. *J Spinal Disord.* 1992;5(4):390-396; discussion 397.
183. Bergmark A. Stability of the lumbar spine. A study in mechanical engineering. *Acta Orthop Scand Suppl.* 1989;230:1-54.
184. Crisco JJ, Panjabi MM, Yamamoto I, Oxland TR. Euler stability of the human ligamentous lumbar spine. Part II: Experiment. *Clinical biomechanics.* 1992;7(1):27-32.

185. Akuthota V, Nadler SF. Core strengthening. *Arch Phys Med Rehabil.* 2004;85(3 Suppl 1):S86-92.
186. Behm DG, Anderson K, Curnew RS. Muscle force and activation under stable and unstable conditions. *Journal of Strength & Conditioning Research (Allen Press Publishing Services Inc.).* 2002;16(3):416-422 417p.
187. Imai A, Kaneoka K, Okubo Y, et al. Trunk muscle activity during lumbar stabilization exercises on both a stable and unstable surface. *The Journal Of Orthopaedic And Sports Physical Therapy.* 2010;40(6):369-375.
188. Hodges PW, Richardson CA. Inefficient muscular stabilization of the lumbar spine associated with low back pain. A motor control evaluation of transversus abdominis. *Spine.* 1996;21(22):2640-2650.
189. McGill SM. Low back stability: from formal description to issues for performance and rehabilitation. / Maintien du bas du dos: de la description formelle au pragmatisme dans un cadre de performance et de reeducation. *Exercise & Sport Sciences Reviews.* 2001;29(1):26-31.
190. Leetun DT, Ireland ML, Willson JD, Ballantyne BT, Davis IM. Core stability measures as risk factors for lower extremity injury in athletes. *Medicine and science in sports and exercise.* 2004;36(6):926-934.
191. Cresswell AG, Oddsson L, Thorstensson A. The influence of sudden perturbations on trunk muscle activity and intra-abdominal pressure while standing. *Exp Brain Res.* 1994;98(2):336-341.
192. Macintosh JE, Bogduk N, Gracovetsky S. The biomechanics of the thoracolumbar fascia. *Clinical biomechanics.* 1987;2(2):78-83.
193. Hodges PW, Richardson CA. Contraction of the abdominal muscles associated with movement of the lower limb. *Phys Ther.* 1997;77(2):132-142; discussion 142-134.
194. Essendrop M, Schibye B. Intra-abdominal pressure and activation of abdominal muscles in highly trained participants during sudden heavy trunk loadings. *Spine.* 2004;29(21):2445-2451.
195. Harman EA, Frykman PN, Clagett ER, Kraemer WJ. Intra-abdominal and intra-thoracic pressures during lifting and jumping. *Medicine and science in sports and exercise.* 1988;20(2):195-201.
196. Kawabata M, Kagaya Y, Shima N, Nishizono H. CHANGES IN INTRA-ABDOMINAL PRESSURE AND TRUNK ACTIVATION DURING DROP JUMP. *Japanese Journal of Physical Fitness & Sports Medicine.* 2008;57(2):225-233.
197. Urquhart DM, Barker PJ, Hodges PW, Story IH, Briggs CA. Regional morphology of the transversus abdominis and obliquus internus and externus abdominis muscles. *Clinical biomechanics.* 2005;20(3):233-241.
198. Hodges PW, Richardson CA. Feedforward contraction of transversus abdominis is not influenced by the direction of arm movement. *Exp Brain Res.* 1997;114(2):362-370.
199. Hodges PW. Changes in motor planning of feedforward postural responses of the trunk muscles in low back pain. *Experimental Brain Research.* 2001;141(2):261-266.
200. Inglin B, Woollacott M. Age-related changes in anticipatory postural adjustments associated with arm movements. *Journal Of Gerontology.* 1988;43(4):M105-M113.
201. Marshall P, Murphy B. The validity and reliability of surface EMG to assess the neuromuscular response of the abdominal muscles to rapid limb movement. *Journal of Electromyography and Kinesiology.* 2003;13(5):477-489.

202. Suri P, Kiely DK, Leveille SG, Frontera WR, Bean JF. Increased Trunk Extension Endurance Is Associated With Meaningful Improvement in Balance Among Older Adults With Mobility Problems. *Archives of Physical Medicine & Rehabilitation.* 2011;92(7):1038-1043 1036p.
203. Pfeifer M, Begerow B, Minne HW, et al. Vitamin D status, trunk muscle strength, body sway, falls, and fractures among 237 postmenopausal women with osteoporosis. *Experimental And Clinical Endocrinology & Diabetes: Official Journal, German Society Of Endocrinology [And] German Diabetes Association.* 2001;109(2):87-92.
204. Suri P, Kiely DK, Leveille SG, Frontera WR, Bean JF. Trunk muscle attributes are associated with balance and mobility in older adults: a pilot study. *PM & R: The Journal Of Injury, Function, And Rehabilitation.* 2009;1(10):916-924.
205. Kahle N, Tevald MA. Core Muscle Strengthening's Improvement of Balance Performance in Community-Dwelling Older Adults: A Pilot Study. *Journal of Aging and Physical Activity.* 2014;22(1):65-73.
206. Teyhen DS, Gill NW, Whittaker JL, Henry SM, Hides JA, Hodges P. Rehabilitative ultrasound imaging of the abdominal muscles. *J Orthop Sports Phys Ther.* 2007;37(8):450-466.
207. Hicks GE, Simonsick EM, Harris TB, et al. Trunk muscle composition as a predictor of reduced functional capacity in the Health, Aging and Body Composition Study: the moderating role of back pain. *Journals of Gerontology Series A: Biological Sciences & Medical Sciences.* 2005;60A(11):1420-1424 1425p.
208. Rankin G, Stokes M, Newham DJ. Abdominal muscle size and symmetry in normal subjects. *Muscle Nerve.* 2006;34(3):320-326.
209. Ikezoe T, Mori N, Nakamura M, Ichihashi N. Effects of age and inactivity due to prolonged bed rest on atrophy of trunk muscles. *European Journal Of Applied Physiology.* 2012;112(1):43-48.
210. Ota M, Ikezoe T, Kaneoka K, Ichihashi N. Age-related changes in the thickness of the deep and superficial abdominal muscles in women. *Archives Of Gerontology And Geriatrics.* 2012;55(2):e26-e30.
211. Brooks GA, Fahey TD, Baldwin KM, eds. *Exercise Physiology: Human Bioenergetics and Its Applications.* 4th ed2005.
212. Enoka RM. *Neuromechanics of human movement* 3rd ed. Champaign, IL: Human Kinetics; 2002.
213. McGill S, Juker D, Kropf P. Appropriately placed surface EMG electrodes reflect deep muscle activity (psoas, quadratus lumborum, abdominal wall) in the lumbar spine. *Journal of biomechanics.* 1996;29(11):1503-1507.
214. Trombetti A, Reid KF, Hars M, et al. Age-associated declines in muscle mass, strength, power, and physical performance: impact on fear of falling and quality of life. *Osteoporosis International: A Journal Established As Result Of Cooperation Between The European Foundation For Osteoporosis And The National Osteoporosis Foundation Of The USA.* 2015.
215. Bischoff-Ferrari H, Orav J, Kanis J, et al. Comparative performance of current definitions of sarcopenia against the prospective incidence of falls among community-dwelling seniors age 65 and older. *Osteoporosis International.* 2015;26(12):2793-2802.
216. Stevens JA, Mahoney JE, Ehrenreich H. Circumstances and outcomes of falls among high risk community-dwelling older adults. *Injury Epidemiology.* 2014;1(1):1-9.

217. Feldman F, Robinovitch SN. Reducing hip fracture risk during sideways falls: Evidence in young adults of the protective effects of impact to the hands and stepping. *Journal of biomechanics.* 2007;40(12):2612-2618.
218. Burkhart TA, Andrews DM. Kinematics, kinetics and muscle activation patterns of the upper extremity during simulated forward falls. *Journal of electromyography and kinesiology : official journal of the International Society of Electrophysiological Kinesiology.* 2013;23(3):688-695.
219. Anderson GS, Gaetz M, Holzmann M, Twist P. Comparison of EMG activity during stable and unstable push-up protocols. *European Journal of Sport Science.* 2013;13(1):42-48.
220. Craig CL, Marshall AL, Sjostrom M, et al. International physical activity questionnaire: 12-country reliability and validity. *Medicine & Science in Sports & Exercise.* 2003;35(8):1381-1395.
221. Monnet T, Desailly E, Begon M, Vallée C, Lacouture P. Comparison of the SCoRE and HA methods for locating in vivo the glenohumeral joint centre. *Journal of biomechanics.* 2007;40(15):3487-3492.
222. Wu G, van der Helm FCT, Veeger HEJD, et al. ISB recommendation on definitions of joint coordinate systems of various joints for the reporting of human joint motion--Part II: shoulder, elbow, wrist and hand. *Journal of biomechanics.* 2005;38(5):981-992.
223. Demura S, Miyaguchi K, Aoki H. The difference in output properties between dominant and nondominant limbs as measured by various muscle function tests. *Journal Of Strength And Conditioning Research / National Strength & Conditioning Association.* 2010;24(10):2816-2820.
224. Allison GT, Godfrey P, Robinson G. EMG signal amplitude assessment during abdominal bracing and hollowing. *Journal of electromyography and kinesiology : official journal of the International Society of Electrophysiological Kinesiology.* 1998;8(1):51-57.
225. Daniels L WK. *Muscle Testing-Techniques of Manual Examination.* Vol 7. Philadelphia, PA: W.B. Saunders Co; 2002.
226. Lehman GJ, MacMillan B, MacIntyre I, Chivers M, Fluter M. Shoulder muscle EMG activity during push up variations on and off a Swiss ball. *Dynamic Medicine.* 2006;5:7-7.
227. Raj IS, Bird SR, Shield AJ. Aging and the force-velocity relationship of muscles. *Experimental Gerontology.* 2010;45(2):81-90.
228. Goodpaster BH, Park SW, Harris TB, et al. The loss of skeletal muscle strength, mass, and quality in older adults: the health, aging and body composition study. *The journals of gerontology. Series A, Biological sciences and medical sciences.* 2006;61(10):1059-1064.
229. Hortobágyi T, Zheng D, Weidner M, Lambert NJ, Westbrook S, Houmard JA. The influence of aging on muscle strength and muscle fiber characteristics with special reference to eccentric strength. *The journals of gerontology. Series A, Biological sciences and medical sciences.* 1995;50(6):B399-B406.
230. Roig M, O'Brien K, Kirk G, et al. The effects of eccentric versus concentric resistance training on muscle strength and mass in healthy adults: a systematic review with meta-analysis. *British Journal Of Sports Medicine.* 2009;43(8):556-568.
231. Pousson M, Lepers R, Van Hoecke J. Changes in isokinetic torque and muscular activity of elbow flexors muscles with age. *Experimental Gerontology.* 2001;36(10):1687-1698.

232. Nagai K, Yamada M, Tanaka B, et al. Effects of balance training on muscle coactivation during postural control in older adults: a randomized controlled trial. *The journals of gerontology. Series A, Biological sciences and medical sciences.* 2012;67(8):882-889.
233. Sale DG. Neural adaptation to resistance training. *Medicine and science in sports and exercise.* 1988;20(5 Suppl):S135-145.
234. Yong Soo K, Yong Ho CHO, Ji Won P. Changes in the Activities of the Trunk Muscles in Different Kinds of Bridging Exercises. *Journal of Physical Therapy Science.* 2013;25(12):1609-1612.
235. Oliver GD, Di Brezzo R. Functional balance training in collegiate women athletes. *Journal Of Strength And Conditioning Research / National Strength & Conditioning Association.* 2009;23(7):2124-2129.
236. Koh TJ, Grabiner MD, Weiker GG. Technique and ground reaction forces in the back handspring. *American Journal of Sports Medicine.* 1992;20(1):61-66 66p.
237. Oliver GD, Plummer HA, Keeley DW. Muscle activation patterns of the upper and lower extremity during the windmill softball pitch. *Journal Of Strength And Conditioning Research / National Strength & Conditioning Association.* 2011;25(6):1653-1658.
238. Allison GT, Morris SL, Lay B. Feedforward responses of transversus abdominis are directionally specific and act asymmetrically: implications for core stability theories. *The Journal Of Orthopaedic And Sports Physical Therapy.* 2008;38(5):228-237.
239. L.J Lattimer JLL, J.P Farthing, S.Y. Kim, S. Madill, S.N. Robinovitch, C.M Arnold. Differences between Younger and Older Women in Muscle Strength and Biomechanics during a Controlled Descent on Outstretched Arms *Canadian Association of Gerentology Annual Conference.* 2014.
240. Hermens BF, R Merletti, G Hägg, D Stegeman, J Blok (Eds.), *et al. SENIAM 8: European recommendations for surface electromyography.* Roessingh Research and Development bv; 1999.
241. Freeman S, Karpowicz A, Gray J, McGill S. Quantifying muscle patterns and spine load during various forms of the push-up. *Medicine & Science in Sports & Exercise.* 2006;38(3):570-577.
242. Florence Peterson Kendall EKM, Patricia Geise Provance, Mary McIntyre Rodgers, William Anthony Romani. *Muscles, testing and function with Posture and Pain.* 5th ed: Lippincott Williams & Wilkens; 2005.
243. Santana JC, Vera-Garcia FJ, McGill SM. A kinetic and electromyographic comparison of the standing cable press and bench press. *Journal Of Strength And Conditioning Research / National Strength & Conditioning Association.* 2007;21(4):1271-1277.
244. García-Vaquero MP, Moreside JM, Brontons-Gil E, Peco-González N, Vera-Garcia FJ. Trunk muscle activation during stabilization exercises with single and double leg support. *Journal of Electromyography and Kinesiology.* 2012;22(3):398-406.
245. Vera-Garcia FJ, Moreside JM, McGill SM. MVC techniques to normalize trunk muscle EMG in healthy women. *Journal of electromyography and kinesiology : official journal of the International Society of Electrophysiological Kinesiology.* 2010;20(1):10-16.
246. Lehman GJ, McGill SM. Quantification of the differences in electromyographic activity magnitude between the upper and lower portions of the rectus abdominis muscle during selected trunk exercises. *Phys Ther.* 2001;81(5):1096-1101.
247. Vera-Garcia FJ, Grenier SG, McGill SM. Abdominal muscle response during curl-ups on both stable and labile surfaces. *Physical Therapy.* 2000;80(6):564-569.

248. Beim GM, Giraldo JL. Abdominal strengthening exercises: A comparitive EMG study. *Journal Of Athletic Training.* 1997;32(2):S31.
249. Hwang JH, Lee YT, Park DS, Kwon TK. Age affects the latency of the erector spinae response to sudden loading. *Clinical biomechanics.* 2008;23(1):23-29 27p.
250. Hodges PW, Eriksson AE, Shirley D, Gandevia SC. Intra-abdominal pressure increases stiffness of the lumbar spine. *Journal of biomechanics.* 2005;38(9):1873-1880.
251. Hodges PW, Gandevia SC, Richardson CA. Contractions of specific abdominal muscles in postural tasks are affected by respiratory maneuvers. *J Appl Physiol.* 1997;83(3):753-760.
252. Tsao H, Hodges PW. Immediate changes in feedforward postural adjustments following voluntary motor training. *Exp Brain Res.* 2007;181(4):537-546.
253. Langevin HM, Fox JR, Koptiuch C, et al. Reduced thoracolumbar fascia shear strain in human chronic low back pain. *BMC Musculoskeletal Disorders.* 2011;12(1):203-203 201p.
254. Canada S. Census of popluation. *Ottawa: Statistics Canada.* Vol Catalogue. no. 98-311-XCB20110172011.
255. Ageing well: a global priority. *Lancet*2012: 1274-1274.
256. Hill C, Riaz M, Mozzam A, Brennen MD. A regional audit of hand and wrist injuries. A study of 4873 injuries. *Journal Of Hand Surgery (Edinburgh, Scotland).* 1998;23(2):196-200.
257. Latash ML, Zatsiorsky VM. Joint stiffness: myth or reality? *Human Movement Science.* 1993;12(6):653-692.
258. Lo J, McCabe GN, DeGoede KM, Okuizumi H, Ashton-Miller JA. On reducing hand impact force in forward falls: results of a brief intervention in young males. *Clinical Biomechanics (Bristol, Avon).* 2003;18(8):730-736.
259. Frost LR, Gerling ME, Markic JL, Brown SHM. Exploring the effect of repeated-day familiarization on the ability to generate reliable maximum voluntary muscle activation. *Journal of Electromyography and Kinesiology.* 2012;22(6):886-892.
260. Oliver GD, Plummer HA, Keeley DW. Muscle activation patterns of the upper and lower extremity during the windmill softball pitch. *Journal of Strength and Conditioning Research.* 2011;25(6):1653-1658.
261. Koyama Y, Kobayashi H, Suzuki S, Enoka RM. Enhancing the weight training experience: A comparison of limb kinematics and EMG activity on three machines. *European Journal Of Applied Physiology.* 2010;109(5):789-801.
262. Shultz R, Silder A, Malone M, Braun HJ, Dragoo JL. Unstable Surface Improves Quadriceps:Hamstring Co-contraction for Anterior Cruciate Ligament Injury Prevention Strategies. *Sports Health: A Multidisciplinary Approach.* 2015;7(2):166-171.
263. Rao G, Amarantini D, Berton E. Influence of additional load on the moments of the agonist and antagonist muscle groups at the knee joint during closed chain exercise. *Journal of Electromyography & Kinesiology.* 2009;19(3):459-466.
264. Mukaka MM. Statistics corner: A guide to appropriate use of correlation coefficient in medical research. *Malawi Medical Journal: The Journal Of Medical Association Of Malawi.* 2012;24(3):69-71.
265. Yu B, Lin C-F, Garrett WE. Lower extremity biomechanics during the landing of a stop-jump task. *Clinical biomechanics.* 2006;21(3):297-305.

266. DeGoede KM, Ashton-Miller JA, Schultz AB, Alexander NB. Biomechanical factors affecting the peak hand reaction force during the bimanual arrest of a moving mass. *Journal Of Biomechanical Engineering.* 2002;124(1):107-112.
267. Jeongok Y, Joongsook L, Bomjin L, et al. The Effects of Elbow Joint Angle Changes on Elbow Flexor and Extensor Muscle Strength and Activation. *Journal of Physical Therapy Science.* 2014;26(7):1079-1082 1074p.
268. Melo RC, Takahashi ACM, Quitério RJ, Salvini TF, Catai AM. Eccentric torque-producing capacity is influenced by muscle length in older healthy adults. *Journal Of Strength And Conditioning Research / National Strength & Conditioning Association.* 2015.
269. Lee Y, Ashton-Miller JA. The effects of gender, level of co-contraction, and initial angle on elbow extensor muscle stiffness and damping under a step increase in elbow flexion moment. *Annals Of Biomedical Engineering.* 2011;39(10):2542-2549.
270. Kuxhaus L, Sisi Z, Robinson CJ. Dependence of elbow joint stiffness measurements on speed, angle, and muscle contraction level. *Journal of biomechanics.* 2014;47(5):1234-1237.
271. SMARTRISK. SMARTRISK: The Economic Burden of Injury in Canada. Toronto2009.
272. Sherrington C, Whitney JC, Lord SR, Herbert RD, Cumming RG, Close JCT. Effective exercise for the prevention of falls: a systematic review and meta-analysis. *Journal of the American Geriatrics Society.* 2008;56.
273. Orr R, Raymond J, Singh MF. Efficacy of progressive resistance training on balance performance in older adults: a systematic review of randomized controlled trials. *Sports Medicine.* 2008;38(4):317-343 327p.
274. Shubert TE. Evidence-Based Exercise Prescription for Balance and Falls Prevention: A Current Review of the Literature...Exercise, Physical Activity and Aging (ExPAAC): Blending Research and Practice, July 29-31, 2010. *Journal of Geriatric Physical Therapy.* 2011;34(3):100-108 109p.
275. Kruse LM, Gray B, Wright RW. Rehabilitation after anterior cruciate ligament reconstruction: a systematic review. *Journal of Bone & Joint Surgery, American Volume.* 2012;94-A(19):1737-1748 1712p.
276. Gentil P, Soares SRS, Pereira MC, et al. Effect of adding single-joint exercises to a multi-joint exercise resistance-training program on strength and hypertrophy in untrained subjects. *Applied Physiology, Nutrition & Metabolism.* 2013;38(6):341-344 344p.

APPENDICES

APPENDIX A. TELEPHONE SCREENING QUESTIONNAIRE

Date:______________

FOOSH Study – Screening Questionnaire

The following questions will provide important information to help us determine if you are eligible for this study. Would you be willing to answer these questions? You can refuse to answer any questions or stop the interview at any time.

1. First, I need to get some contact information.

   Name:
   ____
   ____

   Address:
   ____
   ____

   Postal Code:
   ____
   ____

   Telephone:
   ____
   ____

   e-mail:____

2. What is your age?________________________

3. Have you had any fractures of your wrist/lower forearm? YES☐ NO☐

   If YES, when did it happen? Click here to enter a date. *If it occurred less than 2 years ago, not eligible*
   Which arm RIGHT ☐ LEFT ☐

4. Do you have any previous surgery in your wrists/forearms/spine or legs?

   YES☐ NO☐

   If YES, describe:____

   If they have had any surgery in the arms, check with researcher regarding eligibility.
   Document any other surgeries noted.

5. Have you had any recent (within the past 6 months) injuries to the shoulder/wrist/hand?

YES ☐ NO☐

If YES, describe:____

If recent upper extremity injuries, exclude. If any other injuries noted, check with researcher.

6. Any other recent injuries to other body areas that we should be aware of? Do you have any pain associated with these injuries?

7. Do you have any medical or neurological conditions that you know of.. i.e. peripheral nerve injuries, reflex pain syndrome (arm), stroke, MS..?

   YES☐ NO☐

   If YES, describe:____

   If any current conditions involving weakness or pain in the UE, exclude. If unsure, check with researcher.

8. Do you have any difficulties with balance?

   YES☐ NO☐

   If YES, describe:____

9. Do you have any of the following conditions? If yes, provide details of when diagnosed, current status

Uncontrolled hypertension	YES☐ NO☐	Describe___________
Recent heart attack	YES☐ NO☐	Describe___________
Recent stroke	YES☐ NO☐	Describe___________
Parkinson’s, multiple sclerosis or other Neurological condition	YES☐ NO☐	Describe___________
Congestive heart failure	YES☐ NO☐	Describe___________
Recent lung or blood clot	YES☐ NO☐	Describe___________
Respiratory infection, i.e. pneumonia	YES☐ NO☐	Describe___________
Osteoporosis	YES☐ NO☐	Describe___________
Recent fracture (other than forearm)	YES☐ NO☐	Describe___________
Chest pain/angina	YES☐ NO☐	Describe___________
Vision or Hearing Problems	YES☐ NO☐	Describe___________
Severe arthritis in either wrist or hand	YES☐ NO☐	Describe___________

Any other health problems YES☐ NO☐ Describe___________

If they present with a recent significant medical or neurological concern (i.e. stroke, heart attack, chest pain), inform PI for further follow-up/ and possible exclusion

APPENDIX B. CONSENT FORM 1

PARTICIPANT INFORMATION AND CONSENT FORM

STUDY TITLE Biomechanics and Muscle Activity in Controlled Body Descent Simulating a Fall on the Outstretched Hand in Young and Older Women (FOOSH Study)

PRINCIPAL INVESTIGATOR Dr. Cathy Arnold, Professor School of Physical Therapy, University of Saskatchewan; cathy.arnold@usask.ca or 966-6588

SUB-INVESTIGATORS and/or STUDENT RESEARCHERS
Dr. Joel Lanovaz, Associate Professor, College of Kinesiology
Dr. Jon Farthing, Associate Professor, College of Kinesiology
Lauren Lattimer, PhD student, College of Kinesiology
Matthew Ankerrman, MPT student, School of Physical Therapy
Erin Gibb, MPT student, School of Physical Therapy
Anastasia Slobodzian, MPT student, School of Physical Therapy
Keenan Oberg, MPT student, School of Physical Therapy
Leah Sauchyn, MPT student, School of Physical Therapy
Dr. Stephen Robinovitch, Professor, School of Engineering Science, Simon Fraser University

SPONSOR [or Funding Agency] College of Medicine, University of Saskatchewan, MPT 992 Project Funding

CONTACT NUMBER: 966-8619 OR lauren.lattimer@usask.ca

INTRODUCTION
You are invited to take part in this research study because you are female and fall into one of the following age categories: age 18 – 30 years or age 65 years or older.

Your participation is voluntary. It is up to you to decide whether or not you wish to take part. If you wish to participate, you will be asked to sign this form. If you do decide to take part in this study, you are still free to withdraw at any time and without giving any reasons for your decision.

If you do not wish to participate, you will not lose the benefit of any health care currently being received for any conditions, nor will your academic standing to which you are entitled within the Masters of Physical Therapy program (MPT) be affected. It will not affect your relationship with Dr. Cathy Arnold or any of the MPT student researchers, who may be your peers. Your course instructors will have no knowledge of your involvement in this study, and there is no obligation to participate if you are a physical therapy or kinesiology student.

Please take time to read the following information carefully. You can ask the researcher to explain any words or information that you do not clearly understand. You may ask as many questions as you need. Please feel free to discuss this with your family, friends or family physician before you decide.

WHO IS CONDUCTING THE STUDY?
The study is being conducted by a team of student researchers and other researchers within Kinesiology and Physical Therapy, led by Dr. Cathy Arnold. The Student researchers and an additional research assistant will conduct testing. Funding for the research assistant and equipment is provided by the School of Physical Therapy, College of Medicine, University of Saskatchewan.

WHY IS THIS STUDY BEING DONE?
This study is being done because women are much more likely than men to fracture their wrist, most commonly by **f**alling **o**n **o**ut**s**tretched **h**ands (FOOSH). Although this may be due to a greater loss of bone strength in older women, there are other factors such as muscle strength, reaction time, and body position that may impact the extent of injury sustained in a fall. In other studies comparing younger and older women, older women have had significantly reduced ability to prevent serious injury such as fracture from a forward fall; however the causes of this are unknown. This study will help to determine the causes of differences in lowering the body successfully during a simulated FOOSH in younger and older women, and how muscle activity and performance might contribute to success. This information will be helpful to design intervention programs to decrease the risk of injury from a forward fall.

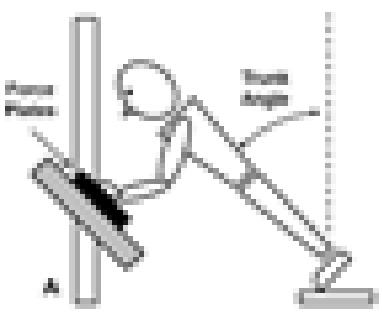
WHO CAN PARTICIPATE IN THE STUDY?
You are eligible to participate in this study if you are female, age 18 – 30 years OR age 65 years and older, with no recent shoulder, arm or hand injuries, any history of severe weakness or instability in the arms or shoulders, i.e. neuromuscular conditions such as peripheral nerve injury, shoulder joint subluxation, regional reflex pain syndrome, etc.

WHAT DOES THE STUDY INVOLVE?
Your involvement will include: 1) a short screening interview conducted by phone or in person, 2) attendance at two testing sessions in the College of Kinesiology that will last 2 – 3 hours in length for each visit. The details of your involvement are outlined below:
<u>Screening Questionnaire</u>: This questionnaire will ask about your age, demographic and contact information, general health and any history of surgery or other conditions that

may affect your ability to participate in this study. If, after completion of this questionnaire, you are eligible for the study, two testing times approximately one week apart will be arranged. You will be asked to wear or bring with you a pair of shorts, and a tight fitting tank top or bathing suit top that exposes the shoulders in order that the markers can be anchored more effectively. If you feel more comfortable wearing a t-shirt, the researchers can accommodate, and if you forget or do not have this type of clothing, there will be a t-shirt and shorts available at the lab. Information and consent forms will be sent or given to you in person prior to testing, and consent forms will be reviewed and signed prior to the testing protocol outlined below:

1) You will complete two questionnaires asking about your physical activity in the past 7 days, and strengthening activities for the arms. You will also complete a handedness questionnaire.
1) Height and Weight will be taken using a standardized ruler on the wall, and weight with a digital weigh scale.

You will then be randomly assigned to either Biomechanical Analysis or Muscle Testing:
Biomechanical Analysis: Approximately 60 reflective markers (spheres) will be placed on your body, including the neck, shoulder, elbows, wrist, a waist band with pelvic marker anchored on it, the knees, feet and on a helmet you will wear over your head (standard hockey helmet with a face guard). The helmet is used only in the rare event that you may have difficulty controlling the descent of your body on the outstretched arms (see figure below), in order to avoid any head impact.



Cameras (motion capture system) around the lab will pick up the spheres' position of your body and will portray a simulated stick figure on a computer screen that will track your movement. The motion capture system cameras only retain the locations of the reflective spheres and no video image is saved. A separate high speed digital video camera will provide a visual reference for subsequent analysis, but these videos are only accessible to the researchers, and have no identification on them other than a subject number. If you consent, by checking the box on the back page, your picture with your identity blocked, may be used for educational purposes. Electromyography (EMG) will be attached over the muscles of your non-dominant upper arm, on your back and abdominal muscles. Surface electrode markers will be used with standard conductive jell and tape. In order ensure the same location of the EMG on the second visit, a piece of plastic wrapping paper will be placed over the body locations noted above temporarily to landmark the location used. Once the markers are secured and the EMG is in place, one of the researchers will resist some movements of your arms and trunk in order to ensure correct placement of the surface electrodes and to normalize the EMG data. You will be asked to contract your muscles as hard as you can against manual resistance, such as flexing your elbow while seated or flexing your upper body while laying on a padded table, holding each contraction for a few seconds, with rest breaks in between. After another short rest period, you will have a harness placed around your waist, with a secure cable attached to a stable hook in the ceiling. You will be in bare feet, and asked to stand on a wooden block anchored to the floor. You will be asked to place both hands on the force plates, and the researchers will set the appropriate height and adjustment of the harness and plates. Two testers will stand alongside in case there is any loss of balance or difficulty with the lowering task. You will first practice slowly lowering yourself using your arm strength, towards the force plate as far as you are able or until the researchers tell you to stop. Once the practice and angle adjustments are complete, you will then complete three trials of descents at four different angles and two different velocities (instructed as slowly lower, and then lower as quickly as you are able to). You will be given two minutes of rest between each change of body position angle. The total testing time will be approximately 1 hour.

Muscle Testing: Specialized equipment called an isokinetic dynamometer will test the strength of your non-dominant arm muscles. You will sit on a secure padded seat, with a harness and waist strap to stabilize your body. Your non-dominant arm will be positioned slightly away from your body with your hand resting on a padded hand rest. You will be asked to do a series of strength tests where you will bend and straighten your elbow and pull your arm against your chest and push outward against resistance with a couple minutes break in between each test. For some of these movements you will be encouraged to push as hard as you can. The test will take approximately one hour to complete.

The same testing protocol will be completed on Day 2, excluding the questionnaires.

WHAT ARE THE BENEFITS OF PARTICIPATING IN THIS STUDY?

If you choose to participate in this study, there may be direct benefits to you such as learning more about your arm strength and ability to lower your body on extended arms. You will also learn about testing methods such as biomechanical analysis, dynamometers, and EMG. It is hoped the information gained from this study can be used in the future to benefit other women, and a copy of the general findings of the study will be sent to you at the conclusion of the study.

ARE THERE POSSIBLE RISKS AND DISCOMFORTS?

If you choose to participate in this study, the following are possible risks and discomforts.

There is the rare risk of injury due to inability to control the descent of the body on the arms. Because this motion will be progressed gradually, first practiced at an angle where you are not supporting your whole body weight (45 degrees), this is extremely unlikely, and the safety precautions to prevent any injury include: 1) a safety harness around the chest and waist, attached to the ceiling, 2) a hockey

helmet with face mask will be worn to avoid any injury to the head or jaw, 3) two spotters will stand beside you at all times and will be close enough to provide support as needed. In the rare event that an injury occurred during the testing session or immediately following, the researchers would arrange for follow-up with your physician or appropriate medical personnel.
You may experience temporary muscular discomfort or joint soreness following the muscle strength testing. This discomfort may last one or two days after the test. Precautions such as ensuring understanding of the protocol, a practice warm-up and close monitoring by the tester will help to alleviate any risks of post-testing discomfort. If you are concerned about any discomfort, please contact the investigators or your family physician.
As with any type of strenuous activity, there is a very small risk that the stress of performing exercise will cause heart rhythm abnormalities, chest discomfort or light headedness. People with a history or presence of significant cardiac (heart) disease or heart rhythm disorders should not participate in this study. The principal investigator may decide that you should not perform the exercise tests, based on information in your medical history or may ask to consult with your family physician before you are accepted in the study. ***It is important that you let the study staff know if you have ever been advised not to participate in strenuous activities***. Minor skin irritation rarely occurs due to the taping of the markers on your skin. If there is some skin itching or redness, it usually disappears in 24 hours.

WHAT HAPPENS IF I DECIDE TO WITHDRAW?
Your participation in this research is voluntary. You may withdraw from this study at any time. You do not have to provide a reason. There will be no penalty or loss of benefits if you choose to withdraw. Your future medical care or academic status will not be affected.

If you choose to enter the study and then decide to withdraw later, all data collected about you during your enrolment will be retained for analysis.

WILL I BE INFORMED OF THE RESULTS OF THE STUDY?
The results of the study will be sent to you from the Principal Investigator, once the study is complete (approximately May – August 2014).

WHAT WILL THE STUDY COST ME?
You will not be charged for any research-related procedures. If you are not already on campus when your testing session is booked, you will receive $5.00 at each visit as compensation for parking if applicable.

WHAT HAPPENS IF SOMETHING GOES WRONG?
By signing this document, you do not waive any of your legal rights.

WILL MY TAKING PART IN THIS STUDY BE KEPT CONFIDENTIAL?
In Saskatchewan, the *Health Information Protection Act (HIPA)* defines how the privacy of your personal health information must be maintained so that your privacy will be respected. Your confidentiality will be respected. No information that discloses your identity will be released or published without your specific consent to the disclosure. However, research records identifying you may be inspected in the presence of the Investigator or his or her designate by representatives of the University of Saskatchewan Research Ethics Board for the purpose of monitoring the research. However, no records, which identify you by name or initials, will be allowed to leave the Investigators' offices. The results of this study may be presented in a scientific meeting or published, but your identity will not be disclosed.

WHO DO I CONTACT IF I HAVE QUESTIONS ABOUT THE STUDY?
If you have any questions or desire further information about this study before or during participation, you can contact the Principal Investigator at 966-6588, the student researcher at lauren.lattimer@usask.ca or contact the Study Phone Line at 966-8619 and one of the researchers will contact you.

If you have any concerns about your rights as a research participant and/or your experiences while participating in this study, contact the Chair of the University of Saskatchewan Research Ethics Board, at 306-966-2975(out of town calls 1-888-966-2975). The Research Ethics Board is a group of individuals (scientists, physicians, ethicists, lawyers and members of the community) that provide an independent review of human research studies. This study has been reviewed and approved on ethical grounds by the University of Saskatchewan Research Ethics Board.

UNIVERSITY OF SASKATCHEWAN

CONSENT TO PARTICIPATE

Study Title: Biomechanics and Muscle Activity in Controlled Body Descent Simulating a Fall on the Outstretched Arms in Young and Older Women

---

- I have read (or someone has read to me) the information in this consent form.
- I understand the purpose and procedures and the possible risks and benefits of the study.
- I was given sufficient time to think about it.
- I had the opportunity to ask questions and have received satisfactory answers.
- I understand that I am free to withdraw from this study at any time for any reason and the decision to stop taking part will not affect my future relationships.
- I give permission to the use and disclosure of my de-identified information collected for the research purposes described in this form.
- I understand that by signing this document I do not waive any of my legal rights.
- I will be given a signed copy of this consent form.
- I am willing to let the researchers contact my family physician if necessary
- I would be willing to be contacted if other research opportunities arise in which I might be eligible

  YES NO
- I would be willing to be photographed for use in educational purposes as long as my identity remains confidential (All identifiers associated with the photograph will be removed (face blocked, a unique code used, and the picture will only include material that does not identify the participant)

  YES NO

I agree to participate in this study:

Printed name of participant: Signature Date

Printed name of person obtaining consent: Signature Date

APPENDIX C. WATERLOO HANDEDNESS QUESTIONNAIRE

Waterloo Handedness Questionnaire

INSTRUCTIONS: Please indicate your hand preference for the following activities by circling the appropriate response. Think about each question. You might try to imagine yourself performing the task in question. Please take your time.

If you use one hand 95% of the time to perform the described activity, then circle right always or left always as your response

If you use one hand about 75% of the time, then circle right usually or left usually.

If you use both hands roughly the same amount of time, then circle equally.

l. Which hand do you use for writing?
Left Always Left usually Equally Right Usually Right Always

2. With which hand would you unscrew a tight jar lid?
Left Always Left usually Equally Right Usually Right Always

3. In which hand do you hold a toothbrush?
Left Always Left usually Equally Right Usually Right Always

4. In which hand would you hold a match to strike it?
Left Always Left usually Equally Right Usually Right Always

5. Which hand would you use to throw a baseball?
Left Always Left usually Equally Right Usually Right Always

6. Which hand do you consider the strongest?
Left Always Left usually Equally Right Usually Right Always

7. With which hand would you use a knife to cut bread?
Left Always Left usually Equally Right Usually Right Always

8. With which hand do you hold a comb when combing your hair?
Left Always Left usually Equally Right Usually Right Always

9. Which hand do you use to manipulate implements such as tools?
Left Always Left usually Equally Right Usually Right Always

| 0. Which hand is the most adept to picking up small objects?
Left Always Left usually Equally Right Usually Right Always

APPENDIX D. INTERNATIONAL PHYSICAL ACTIVITY QUESTIONNAIRE

INTERNATIONAL PHYSICAL ACTIVITY QUESTIONNAIRE

We are interested in finding out about the kinds of physical activities that people do as part of their everyday lives. The questions will ask you about the time you spent being physically active in the **last 7 days**. Please answer each question even if you do not consider yourself to be an active person. Please think about the activities you do at work, as part of your house and yard work, to get from place to place, and in your spare time for recreation, exercise or sport.

Think about all the **vigorous** activities that you did in the **last 7 days**. **Vigorous** physical activities refer to activities that take hard physical effort and make you breathe much harder than normal. Think *only* about those physical activities that you did for at least 10 minutes at a time.

1. During the **last 7 days**, on how many days did you do **vigorous** physical activities like heavy lifting, digging, aerobics, or fast bicycling?

   _____ days per week

   ☐ No vigorous physical activities ➔ ***Skip to question 3***

2. How much time did you usually spend doing **vigorous** physical activities on one of those days?

   _____ hours per day

   _____ minutes per day

   ☐ Don't know/Not sure

Think about all the **moderate** activities that you did in the **last 7 days**. **Moderate** activities refer to activities that take moderate physical effort and make you breathe somewhat harder than normal. Think only about those physical activities that you did for at least 10 minutes at a time.

3. During the **last 7 days**, on how many days did you do **moderate** physical activities like carrying light loads, bicycling at a regular pace, or doubles tennis? Do not include walking.

   _____ days per week

   ☐ No moderate physical activities ➔ ***Skip to question 5***

4. How much time did you usually spend doing **moderate** physical activities on one of those days?

_____ **hours per day**

_____ **minutes per day**

☐ Don't know/Not sure

Think about the time you spent **walking** in the **last 7 days**. This includes at work and at home, walking to travel from place to place, and any other walking that you have done solely for recreation, sport, exercise, or leisure.

5. During the **last 7 days**, on how many days did you **walk** for at least 10 minutes at a time?

_____ **days per week**

☐ No walking ➨ ***Skip to question 7***

6. How much time did you usually spend **walking** on one of those days?

_____ **hours per day**

_____ **minutes per day**

☐ Don't know/Not sure

The last question is about the time you spent **sitting** on weekdays during the **last 7 days**. Include time spent at work, at home, while doing course work and during leisure time. This may include time spent sitting at a desk, visiting friends, reading, or sitting or lying down to watch television.

7. During the **last 7 days**, how much time did you spend **sitting** on a **week day**?

_____ **hours per day**

_____ **minutes per day**

☐ Don't know/Not sure

This is the end of the questionnaire, thank you for participating.

APPENDIX E. CUSTOM BUILT FORCE PLATE APPARATUS

Custom Built Force Plate Apparatus

To safely simulate the different phases of a FOOSH a custom system was built in the Biomechanics of Balance and Movement Lab at the University of Saskatchewan. Previous laboratory simulated experiments used similar methods to support force plates out of the floor [29,41] while other kept the force plates in the floor and horizontally suspended participants over the force plates [218] or had them falling forward from standing [22]. Other laboratory simulations of forward falls did not include specific participant and force plate positioning information in their methodology [40].

This custom built system that rigidly supported two 3D force platforms capable of measuring the forces applied to their surfaces was designed by Dr. Joel Lanovaz. The system allowed for the height and angle of the platforms to be adjusted. The height and angles of the force plates were changed simultaneously and manually by the researchers by using an over head winch (Image 2). An adjustable, industrial-grade safety tether system is incorporated into the ceiling of the testing lab and is secured to a harness worn by the participant (Image 3).

The following images are included in the thesis for the reader to better conceptualize the laboratory methodology in FOOSH 1 and FOOSH 2.

IMAGE 1. Preliminary design (Joel Lanovaz)

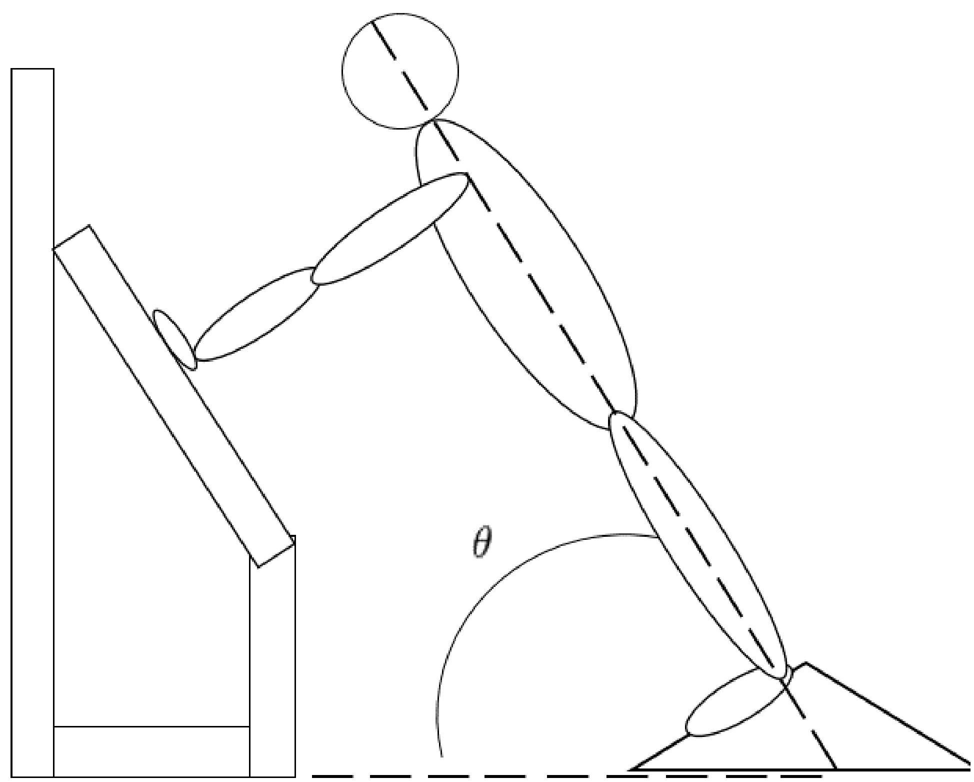
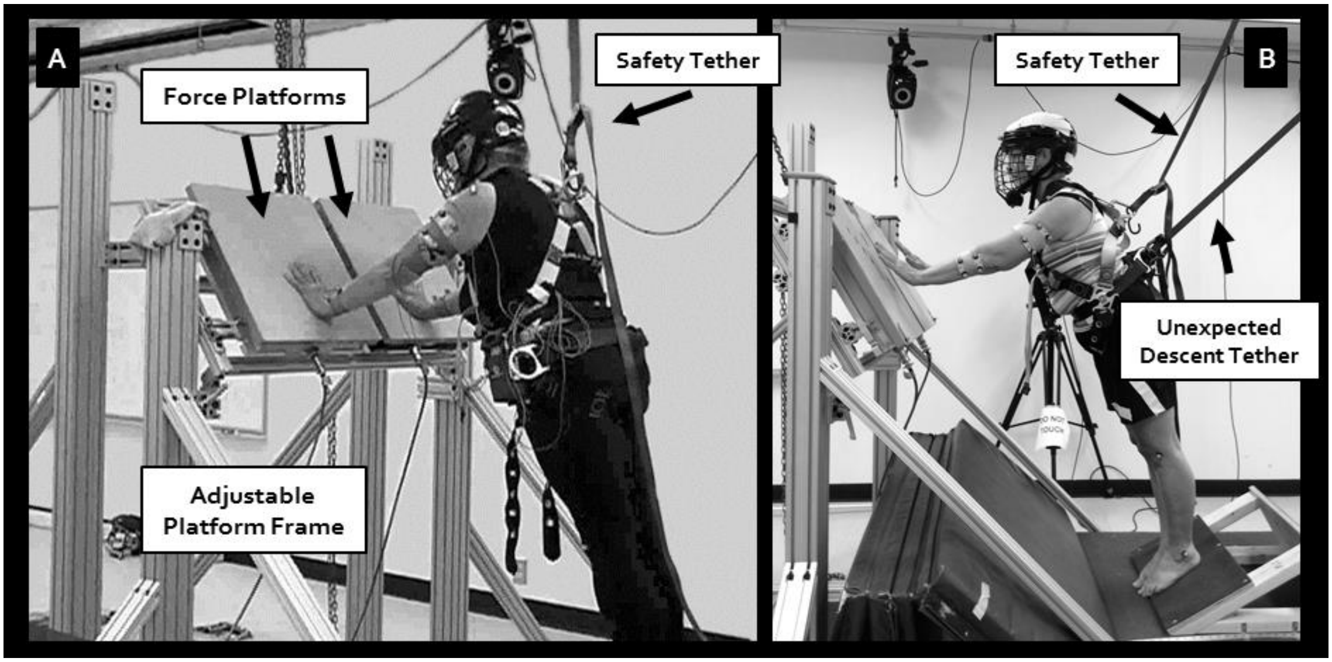


IMAGE 2: FOOSH 1 at body lean angle 60°

Height and angle adjustment winch

IMAGE 3: (A) FOOSH 1 and (B) FOOSH 2



APPENDIX F. COPYRIGHT STUDY 2

RIGHTS AND ACCESS

Help

Article:	Upper limb and trunk muscle activation during an unexpected descent on the outstretched hands in young and older women
Corresponding author:	Ms. Lauren J. Lattimer
E-mail address:	lauren.lattimer@usask.ca
Journal:	Journal of Electromyography and Kinesiology
Our reference:	JJEK1998
PII	S1050-6411(16)30117-1
DOI:	10.1016/j.jelekin.2016.08.001

ORDER SUMMARY

Funding body:	Research reported in the article was not funded by any of the listed funding bodies.
Open Access:	No, I do not want to publish my article gold open access, and would like my final published article to be immediately available to all subscribers.
Publishing Agreement:	I am one author signing on behalf of all co-authors of the manuscript

1st August 2016 T-funding-v4/2010

Privacy Policy Terms & Conditions Help

APPENDIX G. CONSENT FORM 2

PARTICIPANT INFORMATION AND CONSENT FORM

STUDY TITLE Biomechanics and Muscle Activity during an Unexpected Body Descent Simulating a **F**all **o**n the **O**ut**s**tretched **H**and in Young and Older Women (**FOOSH 2 Study)**

PRINCIPAL INVESTIGATOR Dr. Cathy Arnold, Professor School of Physical Therapy, University of Saskatchewan

SUB-INVESTIGATORS and/or STUDENT RESEARCHERS
Dr. Joel Lanovaz, Assistant Professor, College of Kinesiology
Dr. Jon Farthing, Associate Professor, College of Kinesiology
Lauren Lattimer, PhD student, College of Kinesiology

SPONSOR [or Funding Agency]

CONTACT PHONE NUMBER STUDY PHONE LINE 966-8619

INTRODUCTION
You are invited to take part in this research study because you are female and fall into one of the following age categories: age 18 – 30 years or age 60 years or older.

Your participation is voluntary. It is up to you to decide whether or not you wish to take part. If you wish to participate, you will be asked to sign this form. If you do decide to take part in this study, you are still free to withdraw at any time and without giving any reasons for your decision.

Please take time to read the following information carefully. You can ask the researcher to explain any words or information that you do not clearly understand. You may ask as many questions as you need. Please feel free to discuss this with your family, friends or family physician before you decide.

WHO IS CONDUCTING THE STUDY?
The study is being conducted as a part of Lauren Lattimer's PhD thesis. The Student researcher and an additional research assistant will conduct testing.

Funding for the research assistant and equipment is provided by the School of Physical Therapy, College of Medicine, University of Saskatchewan.

WHY IS THIS STUDY BEING DONE?

This study is being done because women are much more likely than men to fracture their wrist, most commonly by falling on an outstretched arm. Although this may be due to a greater loss of bone strength in older women, there are other factors such as muscle strength, reaction time, and body position that may impact the degree of injury sustained in a fall. In other studies comparing younger and older women, older women have had significantly reduced ability to absorb the energy of a forward fall; however the causes of this are unknown. This study will help to determine the causes of differences in lowering the body successfully on outstretched arms and how muscle activity and performance might contribute to success. This information will be helpful to design intervention programs to decrease the risk of injury from a forward fall.

WHO CAN PARTICIPATE IN THE STUDY? (if applicable)

You are eligible to participate in this study if you are female, age 18 – 30 years OR age 60 years and older, with no recent shoulder, arm or hand injuries, any history of severe weakness or instability in the arms or shoulders, i.e. neuromuscular conditions such as peripheral nerve injury, shoulder joint subluxation, regional reflex pain syndrome, etc.

WHAT DOES THE STUDY INVOLVE?

Your involvement will include: 1) a short screening interview conducted by phone or in person, 2) attendance at one or two testing sessions in the College of Kinesiology that will last 1-2 hours in length for each visit. The details of your involvement are outlined below:

Screening Questionnaire: This questionnaire will include questions about your age, demographic and contact information, general health and any history of surgery or other conditions that may affect your ability to participate in this study. If, after completion of this questionnaire, you are eligible for the study, a testing session will be arranged. If after this first session, if you are willing to return for a second test approximately 3 – 5 days apart will be arranged. You will be asked to wear or bring with you a pair of shorts, and a tight fitting tank top or bathing suit top that exposes the shoulders in order that the markers can be anchored more effectively. If you feel more comfortable wearing a t-shirt, the researchers can accommodate.

Each testing session will take approximately 1-2 hours. Information and consent forms will be sent or given to you in person prior to testing, and informed consent will be obtained prior to the testing protocol outlined below:

1) Height and Weight will be taken using a standardized ruler on the wall, and weight with a digital weigh scale.
2) You will complete two questionnaires asking about your physical activity in the past 7 days, and strengthening activities for the arms. You will also complete a handedness questionnaire.
3) You will then do **Muscle Testing** followed by **Biomechanical Analysis.**

2) **Muscle Testing**: Specialized equipment called an isokinetic dynamometer will test the strength of your non-dominant arm muscles. You will sit on a secure padded seat, with a harness and waist strap to stabilize your body. Your non-dominant arm will be positioned slightly away from your body with your hand resting on a padded hand rest. You will be asked to do a series of strength tests where you will bend and straighten your elbow and pull your arm against your chest and push outward against resistance with a couple minutes break in between each test. For some of these movements you will be encouraged to push as hard as you can. The test will take approximately 45 minutes to complete

3) **Biomechanical Analysis:** Approximately 38 reflective markers (spheres) will be placed on select surface markings on the body, including the neck, shoulder, elbows, wrist, a waist band with pelvic marker anchored on it, the knees, feet and on a helmet you will wear over your head (standard hockey helmet with a face guard). The helmet is used only in the rate event that you may have difficulty controlling the descent of your body on the outstretched arms, in order to avoid any head impact. Electromyography (EMG) will be attached over the muscle belly of biceps and triceps on both upper arms. Surface electrode markers will be used with standard conductive jell and tape. In order ensure the same location of the EMG on the second visit, a piece of plastic wrapping paper will be placed over your arm temporarily to landmark the location used. Once the markers are secured and the EMG is in place, a safety harness (similar to what construction workers use) will be placed and secured around your waist and shoulders. You will then be taken to a spot against a wall to test quick arm reaction time by quickly reaching one or both hands to a spot on the wall. An auditory and light cue will signal when to reach. Following this, you will have a chance to practice a quick body release where your hands are just touching the wall with fingertips, and then you quickly lower your body, by placing weight through your hands, lowering as far as able to. If you are comfortable doing this, you will move to stand on a large carpeted

platform attached to the floor, with your safety harness tethered to a ceiling support, facing the force platform. You will be in bare feet, and will stand a set distance away from two force plates elevated and anchored on two support rails. You will be asked to hover your hands on the force plates with just your fingertips touching, and the researchers will set the appropriate height and adjustment of the harness and plates. The research assistant will stand alongside in case there is any loss of balance or difficulty with the task. Your safety harness will have a release tether that will release you from your leaning position forward toward the force plates. The time you will be released will be unknown to you and randomly selected between 1-5sec after a verbal warning. You will then complete five trials of unexpected descents at a 45° angle. You will be given one minutes of rest between each trial. Hovering your fingers lightly on the force plate in the same way as against the wall, you will lower your body by bearing weight through your hands as soon as your body is released. The harness will catch you and not allow you to fall if you have any difficulty controlling your weight through your hands. Cameras (motion capture system) around the lab will pick up the spheres' position of your body and will portray a simulated stick figure on a computer screen that will track your movement. The motion capture system cameras only retain the locations of the reflective spheres and no video image is saved. A separate high speed digital video camera will provide a visual reference for subsequent analysis, but these videos are only accessible to the researchers, and have no identification on them other than a subject number.

The total testing time will be approximately 1-2 hours.

If the participant is able to return for a second testing session, only Biomechanical Testing will be completed.

WHAT ARE THE BENEFITS OF PARTICIPATING IN THIS STUDY?

If you choose to participate in this study, there may be direct benefits to you such as learning more about your arm strength and ability to lower your body on extended arms. You will also learn about testing methods such as biomechanical analysis, dynamometers, and EMG. It is hoped the information gained from this study can be used in the future to benefit other women, and a copy of the general findings of the study will be sent to you at the conclusion of the study.

ARE THERE POSSIBLE RISKS AND DISCOMFORTS?

If you choose to participate in this study, the following are possible risks and discomforts.

There is the rare risk of injury due to inability to control the descent of the body on the arms. Because this motion will be progressed gradually, first practiced at an angle where you are not supporting your whole body weight (45 degrees), this is extremely unlikely, and the safety precautions to prevent any injury include: 1) a safety harness around the chest and waist, attached to the ceiling, 2) a hockey helmet with face mask will be worn to avoid any injury to the head or jaw, 3) a spotter will stand beside you at all times and will be close enough to provide support as needed.
You may experience temporary muscular discomfort or joint soreness following the muscle testing. This discomfort may last one or two days after the test. Precautions such as ensuring understanding of the protocol, a practice warm-up and close monitoring by the tester will help to alleviate any risks of post-testing discomfort. If you are concerned about any discomfort, please contact the investigators. As with any type of strenuous activity, there is a very small risk that the stress of performing exercise will cause heart rhythm abnormalities, chest discomfort or light headedness. People with a history or presence of significant cardiac (heart) disease or heart rhythm disorders should not participate in this study. The principal investigator may decide that you should not perform the exercise tests, based on information in your medical history or may ask to consult with your family physician before you are accepted in the study. ***It is important that you let the study staff know if you have ever been advised not to participate in strenuous activities***. Minor skin irritation rarely occurs due to the taping of the markers on your skin. If there is some skin itching or redness, it usually disappears in 24 hours.

WHAT HAPPENS IF I DECIDE TO WITHDRAW?
Your participation in this research is voluntary. You may withdraw from this study at any time. You do not have to provide a reason. There will be no penalty or loss of benefits if you choose to withdraw. Your future medical care or academic status will not be affected.

If you choose to enter the study and then decide to withdraw later, all data collected about you during your enrolment will be retained for analysis.

WILL I BE INFORMED OF THE RESULTS OF THE STUDY?
The results of the study will be sent to you from the Principal Investigator, once the study is complete (approximately November-December 2014).

WHAT WILL THE STUDY COST ME?
You will not be charged for any research-related procedures. Other than a reimbursement of 5.00 to compensate for parking if applicable, you will not be paid for participating in this study. You will not receive any compensation, or

financial benefits for being in this study, or as a result of data obtained from research conducted under this study.

WHAT HAPPENS IF SOMETHING GOES WRONG?

By signing this document, you do not waive any of your legal rights.

WILL MY TAKING PART IN THIS STUDY BE KEPT CONFIDENTIAL?

In Saskatchewan, the *Health Information Protection Act (HIPA)* defines how the privacy of your personal health information must be maintained so that your privacy will be respected.

Your confidentiality will be respected. No information that discloses your identity will be released or published without your specific consent to the disclosure. However, research records identifying you may be inspected in the presence of the Investigator or his or her designate by representatives of the University of Saskatchewan Research Ethics Board for the purpose of monitoring the research. However, no records, which identify you by name or initials, will be allowed to leave the Investigators' offices. The results of this study may be presented in a scientific meeting or published, but your identity will not be disclosed.

WHO DO I CONTACT IF I HAVE QUESTIONS ABOUT THE STUDY?

If you have any questions or desire further information about this study before or during participation, you can contact the Principal Investigator at 966-6588 or contact the Study Phone Line at 966-8619 and one of the researchers will contact you.

If you have any concerns about your rights as a research participant and/or your experiences while participating in this study, contact the Chair of the University of Saskatchewan Research Ethics Board, at 306-966-2975(out of town calls 1-888-966-2975). The Research Ethics Board is a group of individuals (scientists, physicians, ethicists, lawyers and members of the community) that provide an independent review of human research studies. This study has been reviewed and approved on ethical grounds by the University of Saskatchewan Research Ethics Board.

UNIVERSITY OF SASKATCHEWAN

CONSENT TO PARTICIPATE

Study Title: Biomechanics and Muscle Activity in an Unexpected Body Descent Simulating a Fall on the Outstretched Arms in Young and Older Women

---

- I have read (or someone has read to me) the information in this consent form.
- I understand the purpose and procedures and the possible risks and benefits of the study.
- I was given sufficient time to think about it.
- I had the opportunity to ask questions and have received satisfactory answers.
- I understand that I am free to withdraw from this study at any time for any reason and the decision to stop taking part will not affect my future relationships.
- I give permission to the use and disclosure of my de-identified information collected for the research purposes described in this form.
- I understand that by signing this document I do not waive any of my legal rights.
- I will be given a signed copy of this consent form.
- I am willing to let the researchers contact my family physician if necessary
- I would be willing to be contacted if other research opportunities arise in which I might be eligible
  YES NO
- I would be willing to be photographed for use in educational purposes as long as my identity remains confidential
  YES NO

I agree to participate in this study:

Printed name of participant: Signature Date

Printed name of person obtaining consent: Signature Date